Serratos Pérez, J. Nieves (1994) Postural fatigue of the shoulder: relationships between maximum endurance, subjective perception and electromyographic responses. PhD thesis, University of Nottingham.

Access from the University of Nottingham repository:
http://eprints.nottingham.ac.uk/11540/1/240584.pdf

Copyright and reuse:

The Nottingham ePrints service makes this work by researchers of the University of Nottingham available open access under the following conditions.

- Copyright and all moral rights to the version of the paper presented here belong to the individual author(s) and/or other copyright owners.
- To the extent reasonable and practicable the material made available in Nottingham ePrints has been checked for eligibility before being made available.
- Copies of full items can be used for personal research or study, educational, or not-for-profit purposes without prior permission or charge provided that the authors, title and full bibliographic details are credited, a hyperlink and/or URL is given for the original metadata page and the content is not changed in any way.
- Quotations or similar reproductions must be sufficiently acknowledged.

Please see our full end user licence at:
http://eprints.nottingham.ac.uk/end_user_agreement.pdf

A note on versions:

The version presented here may differ from the published version or from the version of record. If you wish to cite this item you are advised to consult the publisher’s version. Please see the repository url above for details on accessing the published version and note that access may require a subscription.

For more information, please contact eprints@nottingham.ac.uk

POSTURAL FATIGUE OF THE SHOULDER. RELATIONSHIPS BETWEEN MAXIMUM ENDURANCE, SUBJECTIVE PERCEPTION AND ELECTROMYOGRAPHIC RESPONSES

by J.Nieves Serratos Pérez,
Chemical Engineer, Ergonomics MSc

Thesis submitted to the University of Nottingham
for the degree of Doctor of Phylosophy

October, 1994

CONTENTS

page

ABSTRACT ix

ACKNOWLEDGEMENTS x

DEDICATION xi

1. INTRODUCTION 1

1.1 Definition of topic 1
1.2 Background 1
1.3 Postural demands and force exertion as causal factors of the musculoskeletal disorders of the neck and shoulders 3
1.4 The role of the maximum endurance to posture holding 4
1.5 Setting and aims of the investigation 7
1.6 Relevance of the study 7

2. LITERATURE REVIEW 10

2.1 Introduction 10

2.2 Basic issues in muscular fatigue in isometric contraction 11
  2.2.1 Definition of fatigue 11
  2.2.2 Origin and location of fatigue 13

2.3 Cardiovascular responses to isometric exertion 18
  2.3.1 Effect of the strength of the exertion on the blood flow 18
  2.3.2 Changes in heart rate and blood pressure in response to the strength of the exertion 21
  2.3.3 Relationship between the size of the active muscle mass and the cardiovascular responses 22
  2.3.4 Cardiovascular responses to postural efforts 24

2.4 Detection of fatigue using electromyography 27
  2.4.1 Electromyographical signs of fatigue and their likely causes 27
    2.4.1.1 Reduction in the conduction velocity 28
    2.4.1.2 Recruitment of additional motor units 29
    2.4.1.3 Motor unit synchronisation 29
    2.4.1.4 Metabolic factors and muscle fibre type 30
  2.4.2 Considerations to the use of electromyography during isometric exertion 31

page

2.5 Subjective assessment of fatigue and exhaustion in isometric exertion 33
    2.5.1 Nature of the relationship between subjective assessment and the duration of the isometric exertion 34
        2.5.1.1 Pain ratings during continuous handgrip 34
        2.5.1.2 Pain ratings during load holding 36
        2.5.1.3 Ratings of discomfort and pain from the passive loading of joints 38
        2.5.1.4 Discomfort ratings whilst enduring postural loads 38
    2.5.2 Considerations to the linearity of the relationship between subjective ratings and duration of isometric exertion 39
    2.5.3 Issues concerning the application of subjective ratings 43
        2.5.3.1 Conceptual considerations 43
        2.5.3.2 Practical considerations 45

2.6 Postural loads and their effects 48
    2.6.1 'Good' posture versus working posture 48
    2.6.2 Undesirable effects of improper working postures 49
    2.6.3 'Bad' posture as a precursor of musculoskeletal symptoms 50
    2.6.4 Relation between workplace layout, postural loads and musculoskeletal disorders 52
    2.6.5 Absence of a link between postural loads and musculoskeletal disorders 54

2.7 Main issues in work-related musculoskeletal disorders 55
    2.7.1 Factors associated with disorders of the neck and upper limbs 56
    2.7.2 Relationship between occupational factors and disorders of the soft tissues in the shoulder 59
    2.7.3 Ergonomic factors associated with musculoskeletal disorders of the upper body 61
        2.7.3.1 Work mechanisation and static loading of the muscles 61
        2.7.3.2 Physical demands of the task, extreme postures and individual characteristics of the worker 62

2.8 Arm abduction as an ergonomic stressor 66
    2.8.1 Electromyographic evidence of muscular fatigue provoked by arm abduction 66
    2.8.2 Arm abduction as the cause of localised discomfort 68
    2.8.3 Arm abduction as a factor in a variety of work-related illness 69

page

2.9 Physiological mechanisms involved in the musculoskeletal disorders affecting the shoulder 71
  2.9.1 Immediate reactions to muscular strain on the shoulder 71
  2.9.2 Delayed symptoms in response to heavy shoulder usage 72
  2.9.3 Chronic symptoms as a consequence of long-term repetitive usage 73

2.10 Models of fatigue development and recovery in isometric exertion 75
  2.10.1 Monod's model for the calculation of the limit time 76
  2.10.2 Rohmert's model for the calculation of the maximum exertion time 78
  2.10.3 An exponential model of strength and endurance 80
  2.10.4 Rohmert's model for the calculation of rest allowances 86
  2.10.5 Evidence disputing the existence of a 'non-exhausting' level of force 88
  2.10.6 Differences in fatiguability of muscle groups 91

2.11 Models of fatigue and recovery in purely postural work 93
  2.11.1 A model of recovery as a function of the duration of postural loading 93
  2.11.2 Milner's model of recovery as a function of the maximum endurance to postural loading 97
    2.11.2.1 Attempts at widening the application of Milner's model 101
  2.11.3 A work-rest model for purely postural exertion 102
  2.11.4 Application of the work-rest model in the prevention of excessive postural discomfort 106

2.12 Conclusions 109

3. OBJECTIVES AND METHODOLOGY 115

3.1 Wide-scope objectives 115

3.2 Specific objectives 116

3.3 Outline of the experimental work 117

page

3.4 Methodology 119
  3.4.1 Recruitment of subjects 119
  3.4.2 Experimental postures 121
  3.4.3 Layout of the experimental chamber 126
  3.4.4 Preliminary procedures 131
  3.4.5 Collection of discomfort ratings 133
  3.4.6 Procedures followed in the collection of EMG signals 139

4. TESTING OF MILNER'S MODEL FOR THE PREDICTION OF RECOVERY FROM POSTURAL EXERTION 142

4.1 Introduction 142

4.2 Procedures 144
  4.2.1 Determination of the sites of maximum discomfort 145
  4.2.2 Measurement of the maximum holding time 147

4.3 Experimental design 148

4.4 Results 151
  4.4.1 Maximum holding time and degree of recovery 151
  4.4.2 Discomfort ratings 156

4.5 Discussion 157
  4.5.1 Degree of agreement between predicted recovery and observed recovery 157
  4.5.2 Other significant discrepancies between the present and Milner's studies 162
  4.5.3 Assessment of the exponential term in Milner's model 167
  4.5.4 Further information provided by the discomfort ratings 171

4.6 Conclusions 174

5. THE MAXIMUM HOLDING TIME AS A MEASURE OF ENDURANCE TO POSTURAL LOADING 175

5.1 Introduction 175

5.2 Procedures 177
  5.2.1 Subjects 177
  5.2.2 Methods 179

page

5.3 Results and statistical analysis 180
  5.3.1 Length of the maximum holding time 180
  5.3.2 Effects on the MHT of the variation in the posture and the gender of the subject 182
  5.3.3 Consistency of the MHT throughout repeated measures 184
  5.3.4 Influence of the experimental design and the individual characteristics on the variability of the maximum holding time for postural loading 189

5.4 Discussion 198
  5.4.1 The consistency of the maximum holding time 198
  5.4.2 Main influences on the MHT for postural loading 199
  5.4.3 A model to explain the variation of the maximum holding time 201

5.5 Conclusions 204

6. RELATIONSHIP BETWEEN THE MAXIMUM HOLDING TIME AND THE PERCEPTION OF DISCOMFORT 205

6.1 Introduction 205

6.2 Experimental procedure 205

6.3 Results 207
  6.3.1 Background 207
  6.3.2 Sites of unbearable discomfort 208
  6.3.3 The time course of the discomfort ratings 209

6.4 Discussion 216
  6.4.1 Nature of the relationship between the holding time and the subjective perception of discomfort 216
  6.4.2 Expression of the relationship between discomfort ratings and the holding time 223
  6.4.3 Influence of the abduction angle and the gender of the subject on the perception of discomfort 227
    6.4.3.1 Effects of the increase of the abduction angle 227
    6.4.3.2 Comparison of the subjective reactions of male and female subjects to postural loading 228
    6.4.3.3 Assessment of the effects of angle and gender considering the trials individually 230
  6.4.4 Ponderation over the adequacy of 10 as the numerical value for the rating of the maximal discomfort 232
  6.4.5 Discomfort ratings at the upper end of the scale 234

page

6.5 Conclusions 237

7. ASSESSMENT OF MUSCULAR FATIGUE MANIFESTED BY THE CHANGES IN THE EMG SIGNAL 238

7.1 Introduction 238

7.2 Procedures 239
- 7.2.1 Selection of muscles 239
- 7.2.2 Collection of EMG signals 240
- 7.2.3 Spectral analysis 241
- 7.2.4 Reference muscular contractions 242 ✓

7.3 Results 248
- 7.3.1 Main features in the change of the EMG activity over the holding time 248
- 7.3.2 Presence of reversed changes in the EMG parameters 253
- 7.3.3 Level of activation of the muscles during posture holding 254

7.4 Statistical analysis
- 7.4.1 Fitting of a model to the development of fatigue 257
- 7.4.2 Influence of gender and body side on the starting values of MPF and RMS amplitude 266
- 7.4.3 Influence of the abduction angle on the EMG features at the start and throughout the holding 271
- 7.4.4 Extent of the Changes in the EMG Signal During the Postural Exertion 276
- 7.4.5 The nature of the relationship between EMG changes and discomfort ratings 278

7.5 Discussion 283
- 7.5.1 Consistency of the experimental conditions 283
- 7.5.2 EMG changes as indicators of muscular fatigue 284
- 7.5.3 Direction of the changes in the EMG parameters 287
- 7.5.4 Influence of the main experimental factors on the electromyographic changes 288
  - 7.5.4.1 Influence of the gender of the subject 288
  - 7.5.4.2 Influence of the abduction angle 291
- 7.5.5 Persistence of EMG signs of muscular fatigue after posture holding 297

7.6 Conclusions 300

page

8. GENERAL DISCUSSION

8.1 Introduction 301

8.2 The non-applicability of Milner's Model for the prediction of recovery in posture holding 303
  8.2.1 Possible reasons for the non-applicability of Milner's Model 303
    8.2.1.1 Differences between the postures 303
    8.2.1.2 Differences between the experimental designs 307
    8.2.1.3 The gender of the subject 309
  8.2.2 Soundness of Milner's underlying assumptions about the model 309
  8.2.3 Conclusions 311

8.3 Consistency of the Endurance to Postural Loading 312
  8.3.1 Identification of the main influences on the endurance to postural loading 312
  8.3.2 Comparability of the results with those from other studies 313
  8.3.3 Factors that might explain the variation in the maximum holding time 320
  8.3.4 Conclusions 322

8.4 The relationship between postural endurance and the perceived growth of discomfort 323
  8.4.1 Nature of the relationship between holding time and discomfort ratings 323
  8.4.2 Use of the relationship between endurance capacity and subjective perception as a modelling tool 327
  8.4.3 Frequency of appearance of maximum discomfort in the muscles studied 333
  8.4.4 Conclusions 337

8.5 Electromyographic manifestations of muscular fatigue 338
  8.5.1 The relationship between EMG changes and the holding time 338
  8.5.2 The extent of the changes in the EMG signal 341
    8.5.2.1 Size of the changes in RMS amplitude 341
    8.5.2.2 Size of the changes in mean power frequency 343
  8.5.3 Likely causes for the change in the EMG parameters 345
  8.5.4 Relationship between EMG changes and subjective perception 348

page

8.5.5 Reversed changes in the EMG parameters and their implications 352
8.5.5.1 Load sharing and the reversed change of RMS amplitude 353
8.5.5.2 Motor units recruitment in the reversed change of MPF 355
8.5.5.3 Possible causes for the reversed changes in the EMG parameters 356
8.5.6 Persistence of the Signs of Muscular Fatigue. 358
8.5.7 Conclusions 362

8.6 Further work 363

9. CONCLUSIONS 367

REFERENCES 371

APPENDICES

Appendix A Information package handed to the subjects and other printed materials used in the experiments 394

Appendix B Relevant statistics of the regression lines fitted to the data of discomfort rating and holding time collected during each of the 89 trials 411

Appendix C Percentage change of MPF and RMS amplitude observed during each of the 89 trials 417

Appendix D Representative plots of the changes in MPF and RMS amplitude 428

Appendix E Publications generated in the course of the study 489

Serratos-Pérez J.N., Haslegrave C.M. (1992)
Modelling fatigue and recovery in working postures
Contemporary Ergonomics 1992, pp 66-71.
Serratos-Pérez J.N., Haslegrave C.M. (1993)
Monitoring fatigue development from static postures taxing the shoulder region.
The Ergonomics of Manual Work, pp 261-264.
Serratos-Pérez J.N., Haslegrave C.M. (1994)
Relationship between subjective perception and EMG signs of muscular fatigue in shoulder-loading postures.
Accepted for presentation at the 12th IEA Congress

ABSTRACT

The present study was undertaken to investigate issues concerned with the endurance to muscular loads created by the holding of static postures, without the presence of any other form of muscular effort. Its main aim was to explore the possibilities for the development of models which are expected to account for the capacity to endure such kind of exertion.

Upright standing postures, with both arms abducted, were held by young male and female subjects (age 18-24 yr.) for as long as they could, until sensations of physical discomfort, rated on a scale with marks between 0 and 10 (Borg, 1982), became unbearable and forced them to abandon the posture.

The study was constituted by two main experimental stages. In the first one, a posture as described above, with arms abducted at 60°, was used to submit to the test a model developed in 1985 by N.P. Milner (Milner's model). Although this model was originally proposed as a means to predict the remaining proportion of the maximum endurance (or 'recovery') left to the subject after a single sequence of work and rest, where work consisted of the holding of a stooped posture, its author affirmed it could be applied to any posture.

The results of the testing, performed on six female subjects, demonstrated that Milner's model cannot predict with reasonable accuracy the 'recovery' for the upright standing posture with abduction of both arms. Apparently, the assumptions made by Milner concerning the relationships between the endurance capacity and the length of work and rest in a stooped posture did not apply to the test posture.

The second experimental stage had three aims. The first was to test the repeatability of the endurance to standing postures with abduction of both arms. The maximum holding time for postures with arms abducted at 30°, 60° and 90° was measured on three occasions on a sample of five male and five female subjects. The maximum holding time for each of the three postures exhibited a wide variability between subjects, but when compared between the repeated measurements, the average value for the whole sample did not exhibit a significant difference. Also, male subjects had, on average, longer holding times than females, but there was a substantial overlap of the individual values.

The second aim was to investigate the pattern of growth of the discomfort ratings over the length of the maximum holding time. This was found to be of a very strong linear nature, evident in all three postures studied and very similar for men and women. The strength and consistency of this relationship suggest that it may be used as a model to predict either the endurance capacity in function of the rate of growth of discomfort ratings, or the degree of discomfort that a certain length of holding will provoke.

The third aim was to establish whether pure postural loads will provoke changes in the myoelectric activity which indicate the presence of heavy localised muscular fatigue. Mean power frequency (MPF) and RMS amplitude of the EMG signal were monitored throughout the 90 trials of posture holding. Significant changes were evident, with MPF decreasing and RMS amplitude increasing in most of the trials. This means that the posture, even at the lowest angle did provoke muscular fatigue. Another finding, rarely documented, was the presence of electromyographic changes that went in the reversed direction, i.e., MPF increased whilst RMS amplitude decreased. Finally, no well-defined pattern could be established for the time course of those changes or for their relationship with the discomfort ratings.

ACKNOWLEDGEMENTS

I wish to express my endless gratitude to Christine Haslegrave, who throughout the performance of the experimental work, and even more during the writing of this thesis, acted as an untiring advisor, always ready to share the wealth of her impressive scientific knowledge and personal wisdom. Surely, when she first agreed to act as my guide through this research project, little she realised how deep she would have to dig into her reserves of patience and self-restraint in order to meet my demands.

It was only too often that I found myself having to turn to my wife, Carmen Negrete-García (M.D.) and ask her to share her extensive knowledge about work physiology. Every time, an enlightening and fruitful discussion followed from my request. The conceptual background for this research and the final shape of the experimental work owe much to such discussions.

I count myself fortunate in having had the opportunity to meet and interact with Professor Nigel Corlett. I turned to him in my moments of despair, both as a researcher and as a person, and he always found the right words to help me see things from a much brighter perspective. To him, all my gratitude and admiration.

Paddy Riley was a priceless source of advice for the statistical processing of the data. Thanks to him, what looked like an amorphous mass of figures turned into a manageable database, revealing fascinating facts.

It was the British Council who opened the doors of England for me. For that, I will always be grateful to them.

Finally, I wish to acknowledge the support of CONACyT, the National Council for Science and Technology in Mexico, who provided the financial backing for this research.

DEDICATION

Humbly, I wish to dedicate this thesis:

To Carmen, because she was there when I needed her support most;

To Javier, because even the smallest of his achievements makes the largest of mine appear minute by comparison;

To Carolina, because her unconditional love has been a constant source of pride and inspiration to me;

To the memory of my father, Baltasar, who taught me that honesty and hard work are all a person needs to achieve even the hardest and remote of goals;

To my mother, Ernestina, whose pride on her son bears no relation to the size of his achievements;

To my sister, Guadalupe, thanking her for the encouragement she has always offered me;

To all my brothers and sisters.

CHAPTER 1

INTRODUCTION

1.1 Definition of topic

Prolonged holding of static postures is an important issue because of the weight of evidence that suggests the existence of a link between postural demands and the appearance of harmful effects on the muscular and skeletal systems (e.g. van Wely, 1970; Grandjean and Hunting, 1977 Maeda, 1977; Westgaard and Aarås, 1984; Aarås and Westgaard, 1987). Although this issue has been extensively studied in the past few years, there is still a number of aspects where further research can improve on the current understanding. This thesis deals mainly with two of them:

a) the role of the demands placed on muscles by the mere fact of holding static postures, with no other form of effort present;

b) the possibility of using the maximal endurance to the purely postural effort as a means to measure and predict muscular fatigue, with a view to developing tools to prevent the eventual appearance of musculoskeletal damage.

1.2 Background

The last few decades have seen important changes affecting an increasing number of people at work. In many instances, the developments in automation and methods of mass-production have brought with them a considerable reduction of the physical demands on the worker. Thus, jobs in which the worker is required to exert only a relatively small force are now commonplace, not only in relation to office tasks, but also in industries with an important assembly component - the production of electric or electronic goods, the manufacture of garments, shoes and furniture, for example.

However, in many cases the easing of the physical toll exacted by a job has been combined with demands to work at high speed, a circumstance that greatly reduces the opportunity for the worker to move around the workplace. Often this has led to situations where the person remains 'tied' to a machine or workstation that, because of its design, forces the adoption of postures that depart from the more natural ones. Numerous studies (van Wely, 1970; Örtengren et al, 1975; Corlett and Bishop, 1976; Kadefors et al, 1976; Boussenna et al, 1982; Andersson and Örtengren, 1984; Westgaard and Aarås, 1984; Westgaard et al, 1986; Aarås and Westgaard, 1987; Westgaard, 1988) have linked these work conditions with the appearance of excessive muscular fatigue, which could eventually lead to harm on the muscular and skeletal systems. Indeed, some of the studies just mentioned were prompted by the fact that, despite the fairly low levels of force involved in their jobs, significant numbers of workers still developed musculoskeletal troubles.

1.3 Postural demands and force exertion as causal factors of the musculoskeletal disorders of the neck and shoulders

Although they may affect the whole body, the muscular efforts created by the postural constraints present in many assembly line jobs appear to have a more severe effect on the upper part of the body, especially the neck and shoulders. In consequence, a large number of studies have looked into work-related musculoskeletal troubles in these body regions. Some of those studies have been conducted in occupational settings (e.g. Jonsson, 1982; Stranden et al, 1983; Christensen, 1986; Svensson et al, 1987; Hansson et al, 1992; Jensen et al, 1993); others have dealt with specific aspects of the problem in a laboratory setting (Jonsson and Hagberg, 1974; Herberts et al, 1980; Hagberg, 1981a; Sigholm et al, 1984; Strasser et al, 1989; Wiker et al, 1989, 1990; Öberg et al, 1990; Mathiassen and Winkel, 1991); and others have reviewed the epidemiological aspects of the problem (Maeda, 1977; Bjelle et al, 1979; Kilbom et al, 1986; Keyserling et al, 1987; Wallace and Buckle, 1987; Sommerich et al, 1993).

However, practically all the studies looking for a relation between postural demands and musculoskeletal injury to the neck and shoulder have paid much attention to the effort performed by the person in response to external demands, be it by applying force onto objects or by manipulating loads. Less consideration has been given to the demands posed only by the need to keep a posture. Consequently, it is yet to be clearly established to what extent the muscular efforts created by the sole holding of a working posture contribute to the development of musculoskeletal complaints.

Research efforts in that direction have gathered momentum in recent years, but it is not an easy task, as Kilbom (1988) made quite clear. She wrote that whilst, for dynamic conditions, it is relatively easy to find a relationship between muscle strength, endurance time and harmful effects on the shoulder and neck regions, in the case of static, postural load, a mechanism to explain the appearance of those effects is still to be found.

1.4 The role of the maximum endurance to posture holding

In 1978, Barbonis took a completely different approach to the problem of pure postural loading. He studied the development of discomfort experienced by a person when asked to hold a series of stooped postures - enduring what he called 'postural work load'- and the length of rest needed by that person to return to a discomfort-free state. He found that, in the conditions of his study, the time to recover from postural work load was influenced mainly by how long the person had spent holding the posture before being asked to rest, and it appeared unnecessary to know the actual load acting on the muscles involved.

Barbonis (1979) also suggested that the knowledge of holding times alone could be enough to develop models for the prediction of the recovery from purely postural loads. These suggestions are especially appealing in the instance of field studies, for they mean the practitioner could rely mainly on direct observation, without the need to use any equipment or to manipulate the subjects under study.

Milner (1985) applied Barbonis's suggestions in a laboratory study on male subjects holding a stooped posture. He measured the proportion of endurance to posture holding remaining after the subjects underwent combinations of holding and rest, their duration calculated as proportions of the maximum holding time (MHT). The relationship between the length of the holding, the rest and the MHT was built into an equation that may be used to predict the remaining endurance. Milner also suggested that the model was valid for postures other than the one he studied.

In a further development, on the basis that the endurance to postural loading and the perceived discomfort are linearly related (particularly at group level), Dul et al (1991) incorporated Milner's model into a 'work-rest model' that aims to determine the frequency and length of rest periods that should be allowed when a person performs a job with an important postural component. Apparently, these authors took for granted that the equation proposed by Milner is in fact valid for many different postures and for any subject, regardless of gender or age.

In a further development, on the basis that endurance to posture holding - as expressed by MHT- and perceived discomfort are linearly related (not for each individual, but at group level), Dul et al (1993) proposed that the duration of a mainly postural effort should be limited by the time it takes for the person to achieve a certain degree of discomfort. The aim of this proposal, that combines the findings of Barbonis (1979) and Milner (1985), is to limit the discomfort created by the holding of static postures, assuming that this in turn should reduce the risk of musculoskeletal injury.

However, the endurance to any sort of effort is clearly an individual trait, and as such it is potentially subject to wide variations not only between different people, but even for the same person under different circumstances. Milner (1985) considered this point, but found that for the conditions of his study, the inter-individual variations did not appear to affect the validity of his model. Douwes and Dul (1993) also addressed the issue, in relation to the 'work-rest model' proposed by Dul et al (1991). Although Douwes and Dul (1993) identified wide variations in the MHT values measured in theirs and several other authors' previous studies, they still asserted that such variations should not affect the validity of the predictions offered by the model. In sharp contrast, Mathiassen and Winkel (1992) criticised severely the 'work-rest model', precisely on the grounds that the variation of endurance between subjects would render the guidelines based on it practically meaningless, and so the model would be of a very limited value when used as a tool to estimate the risk of musculoskeletal disorders.

1.5 Setting and aims of the investigation

From what has been written so far, it is clear that there are two issues relevant to the area of purely postural exertion that still need to be addressed:

i) are the purely postural demands of a magnitude such as to provoke significant muscular discomfort - which is basically a matter of subjective appreciation- and fatigue, which may be assessed objectively by the changes in one or more physiological variables?;

ii) is it possible to gauge the undesirable effects of the purely postural demands by measuring the endurance to their presence?

The investigation reported here is intended first and foremost as a contribution to the widening of the knowledge about those two issues. It also evaluates whether, as it has been proposed elsewhere, the knowledge of the maximal endurance to the holding of a static posture is enough to predict recovery from this kind of exertion. The major emphasis has been placed on learning about the endurance limits to the postural efforts, the muscular responses with the passage of time, and the degree of discomfort and fatigue that comes from holding postures right to those limits.

1.6 Relevance of the study

As mentioned already, neck and shoulders are the site for a high proportion of work-related musculoskeletal complaints. Three main factors have been consistently linked to the appearance of posture-related trouble in these regions:

a) deviation of the upper arm from a neutral position, mainly in abduction or extension (van Wely, 1970; Chaffin, 1973; Keyserling et al, 1987);

b) lack of support for the arms (Schüldt et al, 1985; Serratos-Pérez and Mendiola-Anda, 1993; Schierhout et al, 1993);

c) location of the hands at or above shoulder level (Bjelle et al, 1979; Herberts et al, 1980; Hagberg, 1984; Wiker et al, 1989).

Any combination of these factors will place the muscles of the shoulder region under significant stress.

Significant numbers of workers in a variety of industries are subjected to these conditions, often for a considerable proportion of their work time. In the experience of this researcher, in the shoemaking industry in Mexico alone there are between 100,000 and 150,000 people, most of them sewing machine operators, whose job requires them to work with their arms abducted and without support (Serratos-Pérez and Mendiola-Anda, 1993), and it is reasonable to assume that at least a similar number of workers in the garment-making industry in that country face the same working conditions. This researcher has also observed a series of jobs in the production of knitwear and garments in the East Midlands where the workers (a total of between 5,000 and 10,000 people in the United Kingdom) spend considerable time with at least one arm in abduction that sometimes goes beyond 90°, placing the hand at or above shoulder level (Serratos-Pérez, 1990).

This study considers specifically the following issues, which are highly relevant to the working conditions observed in those jobs:

1) what is the response of the muscles of the shoulder region to the stress of purely postural origin?;

2) could the maximum endurance to this stress provide a reliable means to evaluate muscular fatigue among the workers subjected to it, or is it the case that the the variations between individuals are too large?;

3) is there a relationship between the maximal endurance to posture holding, the development of the subjective perception of discomfort, and other means used for the assessment of fatigue?;

4) can a single model be found, that permits the assessment of postural fatigue and recovery in any posture and for any subject?

The rationale for carrying out this study is that by reaching a better understanding of the role played by postural strain alone in the appearance of excessive discomfort and fatigue, it should then be easier to work out the relevance of this factor in the eventual development of musculoskeletal injury, when other factors such as external loads and force exerted by the person come into play.

CHAPTER 2

LITERATURE REVIEW

2.1 Introduction

Over the past 20-30 years the interest in isometric exertion has grown markedly, the research effort in the area has increased noticeably and consequently the state of the knowledge is rapidly changing. The amount of published research in this area has grown steadily over the past few years, and this makes it necessary to limit this review to those works which are most relevant to the aims of the present study. Where appropriate, the review will delve more deeply into the issues related to the fatigue of the muscles of the shoulder provoked by static abduction, and the effects on those muscles, since such was the central concern of the study.

The review comprises three main subjects. The first of them refers to the wide-ranging issues of fatigue and its manifestations in terms of both physiological and psychophysical changes, which are covered in sections 2 - 5 of the chapter. The second main subject is covered in sections 6 - 9 of the chapter, referring in first instance to the effects on the musculoskeletal system that have been attributed to the adoption of awkward postures, then looking into a variety of work-related and individual factors that have been associated with the appearance of those harmful effects on the shoulder region, concentrating then on the role of shoulder abduction as a specific risk factor, and finally reviewing the physiological mechanisms that are believed to be behind the musculoskeletal disorders that affect the shoulder. Sections 10 and 11 of the chapter cover the third main subject in the review; the first part looks into the work performed in trying to establish what relationship exists between the force involved in an isometric exertion and the length of time it may be sustained, while the second part deals with the attempts at finding a model to express the course of fatigue and recovery in cases of purely postural exertion.

2.2 Basic issues in muscular fatigue in isometric contraction

Since this thesis is concerned with the issue of muscular fatigue, it is convenient to start by reviewing, albeit briefly, the definition and manifestations of this phenomenon.

2.2.1 Definition of fatigue

Practically every attempt at defining fatigue is preceded by a statement about the complexity of the issue and how in the end it becomes necessary to propose a different definition to suit each particular approach to the problem. Thus, De Luca (1985) cites the proposal by Bills (1943) that there should be at least three definitions of fatigue: subjective, manifested as a decline of concentration, motivation and alertness; objective, evidenced by a decline in the work output; and physiological which is characterised by changes in the physiological processes.

In the opening of the Ciba Foundation Symposium 82: 'Human Muscle Fatigue: Physiological Mechanisms', Edwards (1981) mentioned that the word 'fatigue' has many different meanings, and presented a list with some definitions of the term. That list is (verbatim) reproduced below:

Definition

1. Impaired intellectual performance
2. Impaired motor performance
3. Increased EMG activity for given performance
4. Shift of EMG power spectrum to low frequencies
5. Impaired force generation

Confusion of perception associated with fatiguing muscular activity

1. Increased effort of maintaining force
2. Discomfort or pain associated with muscular activity
3. Perceived impairment of force generation

Clearly, definitions 1 and 2 correspond to those proposed by Bills (1943, cited in De Luca, 1985) as definitions of subjective and objective fatigue; definitions 3 to 5 are examples of physiological manifestations of fatigue at muscle level. It is interesting that the second part of the list includes the different ways in which a person perceives the presence of muscular fatigue, albeit under a heading which in itself results rather confusing. More specifically, Edwards (1981) defined muscular fatigue as "a failure to maintain the required or expected force". Jones and Round (1990) rephrased that definition to "muscular fatigue is a loss of the ability to generate force...", adding that even when so defined, it is important to bear in mind that the extent of the fatigue detected may well vary depending on the method used to measure it.

However, De Luca (1985) finds it preferable to think of muscular fatigue in a way similar to that applied in the physical sciences and engineering, where fatigue is considered as a time-dependent process of change that will eventually culminate at a failure point. He contends that defining muscular fatigue as a phenomenon associated with a particular event happening at a certain moment only takes account of the failure point, missing completely the whole process that led to it.

2.2.2 Origin and location of fatigue

If fatigue is defined as the failure by the muscle to generate the required force, this then raises the issue of whether such failure occurs in the portion of the command chain corresponding to the central nervous system (central fatigue) or whether it is due to changes that occur in the muscle itself (peripheral fatigue) and that affect directly the capacity of the muscle fibre to generate the force through the necessary chemical reactions. Jones and Round (1990) have presented a comprehensive review of the knowledge gained in trying to answer those questions; they stressed the fact that most of this knowledge originates from experiments on single muscles that are submitted to isometric contraction at maximum strength, a situation that can only be sustained for about 60 seconds. What follows is a brief summary of their findings.

A major difficulty in trying to understand the origins of fatigue lies with the fact that not even the source for the sense of effort that accompanies any form of muscular exertion has been precisely identified. There are two currents of opinion in this respect: one holds that the motor centre communicates directly with the centre responsible for the sensing of the effort, the other suggests that the information originates at the sensory receptors in the muscle itself. Either way, the activity of the central nervous system involved in the sustained contraction of a muscle is modified in the course of the exertion, and that modification could eventually result in a failure appearing at some point in the pathway connecting the higher nervous centres with the muscle units. There are three locations in that pathway where it is possible to check for the existence of a failure, and these are illustrated in figure 2.1. Checks can be performed in the three locations by inserting an electrical stimulus at those points and observing the response of the muscle.

Jones and Round (1990) found only one case where they judged that the evidence presented by the authors pointed to the existence of fatigue that originated at the level of the higher centres. This was a study of the maximal exertion of quadriceps by Bigland-Ritchie et al (1978, cited in Jones and Round, 1990).

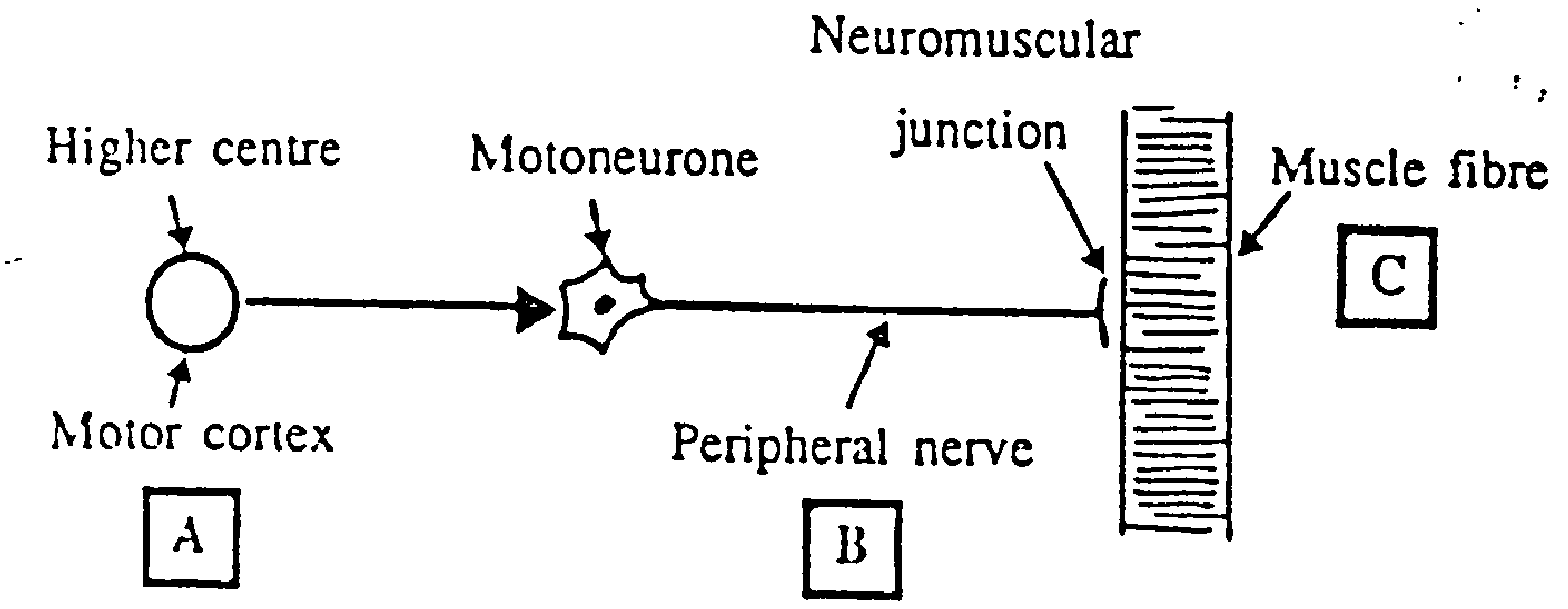


Figure 2.1 The chain of command linking the higher centres with the muscular contraction. The letters A, B, C refer to the sites where electrical stimulation can be used to test the function of the chain. (From Jones and Round, 1990).

The neuromuscular junction is the next location where a failure leading to fatigue could be found. The mechanism for that failure would be the depletion of the stores of acetylcholine (AC) to a level lower than that required for the propagation of the action potential beyond the post-synaptic membrane. The way to check for a failure at this level is to compare the amplitude of the action potential obtained when the muscle is stimulated via the motor nerve (passing through the neuromuscular junction) before the start of the exertion with that of the action potential obtained in the same way during the exertion. A decrease in the amplitude would mean that the neuromuscular junction of a number of muscle fibres had failed, probably due to the exertion. However, this issue has not yet been resolved; whilst there are studies that have found no modification of the action potential (Merton, 1954; Bigland-Ritchie et al, 1982; cited in Jones and Round, 1990) others have produced the opposite evidence (Stephens and Taylor, 1972; cited in Jones and Round 1990).

The remaining possible location of the failure that leads to fatigue is the muscle fibre itself, where the sustained maximal activity provokes changes in the concentration of the substances involved in the liberation of energy, and of the metabolites generated as by-products of the reactions involved. An example is presented in figure 2.2, which shows the changes that occurred during the sustained maximum isometric contraction of the first dorsal interosseus muscle, lasting 45 seconds. The curve in the upper part of the figure shows the reduction of the force generated by the muscle as the contraction time elapsed. The curves in the lower part show the change in the concentration of phosphocreatine (PC), adenosine triphosphate (ATP) and inorganic phosphate (Pi) that accompanied the reduction in force. The numbers between the two portions of the figure show the intracellular pH.

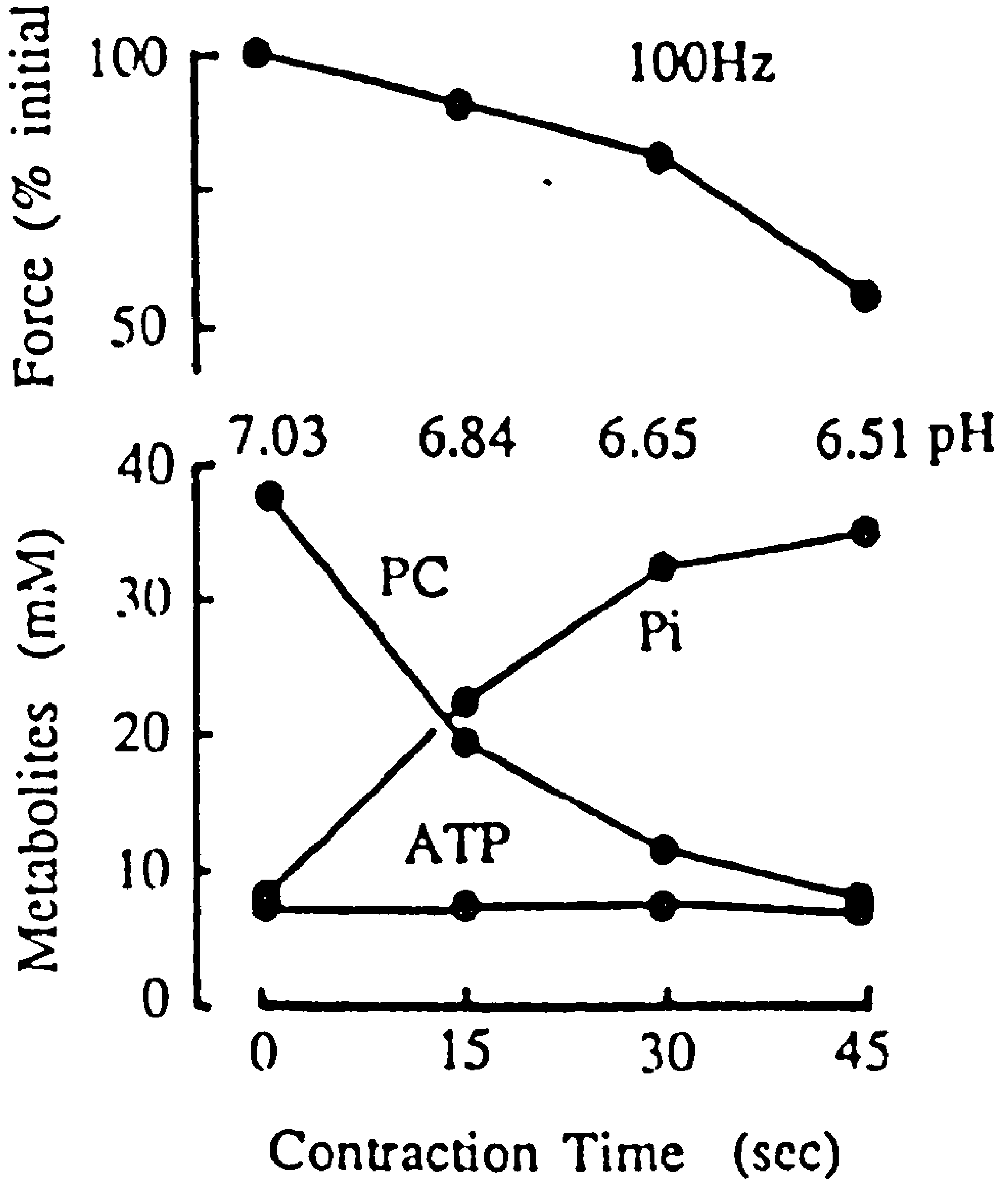


Figure 2.2 Level of force generated and concentration of muscle metabolites during sustained contraction of first dorsal interosseus muscle. (From Jones and Round, 1990).

Figure 2.2 shows that the concentration of phosphocreatine fell markedly throughout the exertion, and this resulted in the concomitant rise of inorganic phosphate. There was a linear increase (not illustrated) in the concentration of lactate during the whole exertion, which was related to the decrease in pH. Finally, the concentration of ATP remained practically constant. According to Jones and Round (1990) these changes relate to the reduction of the force generated by the muscle as follows: since it remains practically unchanged, the concentration of ATP per se is of very little consequence. The fall in pH plays some part in the force reduction, but cannot be singled out as the only cause. The increase in the inorganic phosphate affects the viability of some stages in the process of force production, and in combination with the fall in pH its influence is even larger. However, at the present time it is not known to what extent that influence could explain fatigue.

The remainder of the information presented by Jones and Round (1990) is mainly concerned with the detailed description of the intracellular changes that have been suggested as the most likely mechanisms for the development of fatigue. Since such a level of detail is beyond the scope of this review, that information will not be considered. However, it is worth bringing up the closing comment expressed by those authors, regarding the fact that, despite extensive efforts in the search for an explanation for the appearance of muscular fatigue, this is still far from being properly understood.

2.3 Cardiovascular responses to isometric exertion

The responses of the cardiovascular system to isometric exertion have been extensively studied. It has been established that heart rate, blood pressure and cardiac output increase in response to the onset of isometric effort (Tuttle and Horvath, 1957). It appears that these responses are controlled by a combination of two neural mechanisms. One mechanism is initiated in the higher nervous centres and operates through the central command which drives the cardiovascular system (Krogh and Lindhard, 1913), the other responds to reflexes that originate in the contracting muscles (Paterson, 1928). Although a detailed discussion of these mechanisms is not within the scope of this review, it is convenient to mention that despite having been identified a long time ago, to date their precise roles have not been elucidated beyond doubt.

The remainder of this section will review the relationship between the cardiovascular responses and the main factors present in an isometric muscular contraction, that is the strength applied during the exertion, the muscular mass involved, and the mode of the exertion.

2.3.1 Effect of the strength of the exertion on the blood flow

This area of research has been strongly influenced by the idea put forward by Rohmert (1960), in the sense that a muscular exertion that involved less than 15% of the maximal strength of the muscle (maximal voluntary contraction, or MVC) was practically non-fatiguing and could be sustained for very long times.

Such influence is evident in the comment made by Nutter et al (1972). They wrote that as long as the strength of the isometric exertion is 15% MVC or less, the increase in the blood pressure combined with the vasodilatation that accompanies the exertion will bring enough blood to the muscle to meet the metabolic demands and so avoid fatigue, with the result that an exertion of that intensity might be sustained indefinitely.

However, the results reported by Sjøgaard et al (1988) refute that assertion. Those results were obtained during two series of studies, in which the authors tried to elucidate the role of the reduced blood flow through the muscle in explaining the appearance of fatigue. One series of studies involved the contraction of quadriceps with force equal to 5, 15, 25 and 50% MVC, the last two intensities sustained to exhaustion. The other studies consisted of handgrip contractions where the force applied was 10%, 20% or 40% MVC, the duration of the effort adjusted so that the total amount of work was the same in the three exertions. In both series of studies, exertion intensities higher than 10% MVC provoked reductions of the blood flow which may in fact be seen as an important contributory factor to the appearance of fatigue. Sjøgaard et al (1988) found that when the force applied was 10% MVC or lower, the blood flow through the exercising muscle was at a level compatible with the maintenance of homeostasis; however, the subjects found those exertions fatiguing. Furthermore, when tested following 1 hour of contraction at 5% MVC, they could only produce on average 90% of their initial maximal strength.

Byström and Kilbom (1990) studied the response of the blood flow to continuous as well as intermittent isometric handgrip at intensities of 10, 25 and 40% MVC. The intermittent exercise combined work and rest of duration 10+10, 10+5 and 10+2 seconds. They assessed the acceptability of the work regimes in function of the change in the blood flow through the forearm when the subjects switched from contraction to relaxation, as well as the rating of the perceived effort by the subjects. All three combinations of intermittent work-rest at 10% MVC, and the combinations 10+10 and 10+5 seconds at 25% MVC were deemed acceptable on both accounts. Regarding the continuous exertion, the authors found that 25% and 40% MVC were unacceptable straightaway on both criteria, and although 10% MVC appeared acceptable in terms of the change of blood flow, the subjects perceived the effort to be unacceptably high. This result agreed fully with the findings of Sjøgaard et al (1988).

Nevertheless, the effect of the level of strength used during the exertion on the blood flow through the active muscle has not been established beyond controversy. Thus, in a study of the blood flow during continuous handgrip to exhaustion, Humphreys and Lind (1963) found that blood flow through the exercising muscle increased as the relative strength of the contraction increased, up to a level of 50% MVC; beyond this point, however, the flow decreased until it was practically stopped when the relative strength reached 70% MVC and over. In contrast, Gaffney et al (1990) reported that during the continuous exertion of quadriceps at strength of 15, 25 or 50% MVC, the blood flow through the active muscles decreased instead of increasing, and the effect was more noticeable with the increase of the strength.

2.3.2 Changes in heart rate and blood pressure in response to the strength of the exertion

Rohmert (1960) also suggested that for isometric exertion involving forces below 15% MVC there should be a circulatory steady-state (Sakakibara and Yonda, 1990). This assertion led many researchers to consider only muscle exertions of a strength larger than such 'cut-off' point, as in the studies by Donald et al (1967) and by Lind and McNicol (1967) who reported that in submaximal exertion, when the strength applied was larger than 15-20% MVC, both blood pressure and heart rate increased proportionally to the force applied.

However, Fallentin and Jørgensen (1992) reported findings that contradict the existence of the circulatory steady-state when the force exerted is lower than 15% MVC. They studied the response of the mean arterial blood pressure to elbow flexion or extension sustained to exhaustion whilst applying forces equal to either 10% or 40% MVC. They found that the contraction at 10% MVC provoked a continuous, progressive increase in the mean arterial pressure, so that the terminal value was either equal to that observed at 40% MVC (end point of the elbow extension) or only slightly lower (end point of the elbow flexion). This result showed that the response of the blood pressure to the exertion at the lower intensity was indeed comparable to the effects created by a much higher force. However, the authors emphasised that this similarity of effects between such different loads only became apparent when comparing results from exertions sustained to exhaustion, which for the exertion at the lower force could mean a continuous effort lasting for several hours.

2.3.3 Relationship between the size of the active muscle mass and the cardiovascular responses

It has been reported that in submaximal exertion, when the strength applied was larger than 15-20% MVC, both blood pressure and heart rate increased proportionally to the force applied and the duration of the effort, but were independent of the muscle mass involved in the exertion (Donald et al, 1967; Lind and McNicol, 1967). However, the results reported by Kilbom and Persson (1981) only partially agreed with those earlier reports. They compared the cardiovascular responses elicited by the exertion of three muscle groups at two intensities of exertion. The modes of exertion studied were handgrip (finger flexor muscle), leg extension against resistance (quadriceps muscles) and plantar flexion against resistance (soleus muscle); the three manoeuvres were performed applying 15% MVC for 6 minutes and 25-30% MVC held to exhaustion or for a maximum of 6 minutes. They found that for each muscle group the cardiovascular responses were more pronounced as the intensity of the exertion increased, in line with the findings of Lind and McNicol (1967).

However, Kilbom and Persson (1981) found that the size of the responses was not the same for the three muscle groups at the same intensity of exertion. Thus, the larger increase in heart rate and blood pressure occurred with the contraction of the quadriceps at 25% MVC; it was smaller for the handgrip at 30% MVC and even smaller for the foot plantar flexion at 30% MVC. These results did not exactly mean that the exertion that involved the larger muscle mass provoked the larger effect on the cardiovascular system, since although it was actually the exertion of the larger muscles (quadriceps) that was accompanied by the more pronounced response, it was the muscle of intermediate size (soleus) which provoked the smallest change.

Misner et al (1990) also tested the hypothesis that the larger the muscle mass used in a static exertion, the more pronounced the cardiovascular responses should be. They studied the responses to 2-minute long contractions at maximum strength of the right hand finger flexors (handgrip), right leg quadriceps and both legs' quadriceps (attempted extension against resistance in both cases); they also compared the responses shown by male and female subjects. They found that systolic, diastolic and mean arterial blood pressure all rose continuously in the three forms of exertion. Heart rate, however, only increased continuously during the extension of both legs by the female subjects; in all the other cases it increased at first and then decreased. This kind of response by the heart rate appeared to be unique to their study and Misner et al (1990), attributed it to the fact that their subjects performed a maximal contraction for 2 minutes, as opposed to the 10 seconds that seem to be the typical length of maximal exertion used in similar studies.

Misner et al (1990) reported that the size of the cardiovascular responses was in fact related to the muscle mass involved, so that it was the largest for the exertion with both quadriceps muscles, smaller when only one of them was involved, and the smallest for the exertion of the fingers flexor muscle. They also found that male and female subjects responded in very similar ways to the static exertion, although the males had slightly lower heart rate and higher values of systolic blood pressure during the exertion.

2.3.4 Cardiovascular responses to postural efforts

All the studies mentioned so far have been related to situations where the experimenter determines what muscle group or groups will be submitted to isometric exertion, and the extent of the force that will be used. Mathiassen and Winkel (1991) presented the results from a very interesting study, aimed at comparing the cardiovascular effects of two modes of low-level static contraction where the exertion was actually of postural nature. This is a situation in which although it is possible to identify the muscle group most heavily engaged in the exertion, it is not possible to ensure that no other muscles will be involved, nor is it feasible to quantify precisely the force being exerted.

Mathiassen and Winkel (1991) asked 6 female subjects to perform two experimental protocols in which they were required to hold both arms stretched to the front, at an angle of 60° in the saggital plane. One of the protocols consisted in a single holding of that posture until the subject reached the exhaustion point; the second one combined a holding lasting 300 seconds with a rest period 60 seconds long and the subjects performed as many sequences as they could until they reached exhaustion. Both protocols included the performance of test contractions before the start of the holdings, immediately after they finished, 1 hour and 4 hours after this point. To accomplish the test contraction the subjects adopted the same posture as during the experiments and held for a minute a weight corresponding to 25% of their maximum strength. Heart rate and mean arterial pressure were measured with non-invasive methods both during the exertion and the test contractions.

Both the continuous and the intermittent exertion provoked significant increases in heart rate and mean arterial pressure, as compared with the corresponding values at rest. The mean arterial pressure was significantly higher at the end of the intermittent holdings than at the end of the continuous one, but the heart rate did not show a significant difference at this point. However, in the measurements obtained from the test contractions performed one and four hours after the end of the experiments, both heart rate and mean arterial pressure were significantly higher following the sequential exertion than following the continuous one. This result showed that although both forms of exertion created the same extent of load on the cardiovascular system, the sequential exposure to work and rest had the longer-lasting effects.

Barbonis (1979) conducted a study in which he measured the endurance to a series of five postures in which the hands were located at variable height and distance from the body, defined in relation to the shoulder height and arm reach. He observed the cardiovascular response to those postures by continuous recording of heart rate, finding that there was not a proportional relationship between this variable and the extent of the loads created by each posture, since not always the posture which the subject could endure the least was the one which provoked the larger change in heart rate.

Using a stooped posture taken from those studied by Barbonis (1979), Milner (1985) performed another study of endurance to postural load. In his experiments the subjects underwent a sequence of a submaximal holding time (a time shorter than their endurance limit), followed by rest and then a second holding to the point of exhaustion. During the two stages of postural exertion the heart rate was recorded continuously and the blood pressure was measured on several occasions. Heart rate exhibited a significant linear increase with the passage of the holding time and although both systolic and diastolic pressure showed a linear increase in relation to the holding time, this was significant only for the systolic pressure. This result diverged from the findings of most other studies on isometric exertion.

In a study where the main interest is the endurance to postural exertion, the measurement of blood pressure and heart rate constitutes a distracting factor that might affect the outcome of the work, as Milner (1985) pointed out. This is so mainly because the manoeuvres necessary to obtain the measurements interfere with the subject keeping the desired posture, and the repeated distraction might end up affecting the willingness of the subject to carry on with the exertion until they actually reach the endurance limit.

2.4 Detection of fatigue using electromyography

Although its principles have been well established for almost a century, the collection and analysis of electromyographical information only started to be applied in studies of functional anatomy around the 1940's (Jonsson, 1978). Since then, and particularly over the last three decades, the analysis of electromyographic signals collected from superficial muscles has been extensively used in studies of isometric exertion.

2.4.1 Electromyographical signs of fatigue and their likely causes

A shift towards lower frequencies in the spectrum of the electromyographical signal is a sign of muscular fatigue, as Kogi and Hakamada (1962) were among the first to report, and a large number of later studies have confirmed. Increased amplitude of the signal is another change that has been repeatedly associated with the presence of fatigue (Kadefors et al, 1968; Viitasalo and Komi,1977; Hagberg, 1981a, and others).

The causes behind the spectral alterations that appear with fatigue have been extensively studied and a variety of mechanisms have been proposed, but to date it is not possible to say that the matter has been settled. De Luca (1985) summarised the main explanatory attempts under three headings: a) modification of the conduction velocity of the muscle fibres, b) motor unit recruitment, c) motor unit synchronisation.

2.4.1.1 Reduction in the conduction velocity

Decrease in the conduction velocity of the action potential along the muscle fibre has been proposed as the major contributor to the spectral changes (Lindström et al, 1970, 1977), but this view has been strongly challenged in a series of recent studies that have found only a very limited correspondence between the extent of the spectral change and the conduction velocity (Krogh-Lund and Jørgensen, 1991,1992, 1993; Krogh-Lund, 1993). Another issue apparently settled by this series of studies is that, contrary to what Lindström and Petersén (1983) suggested, electromyography may in fact be used to detect the fatigue provoked by isometric contractions at a force below 20% MVC.

2.4.1.2 Recruitment of additional motor units

Edwards and Lippold (1956) proposed that if a muscle is to generate a constant force during isometric contraction, this will be possible only if additional motor units are constantly being recruited to replace those that have lost their contractility through fatigue, a fact that will be reflected by an increase in the amplitude of the electromyographic signal. Eason (1960), Maton (1981), and Moritani et al (1982) among others, have subscribed to that view. More recently, Arendt-Nielsen et al (1989) and Hägg (1991) have also invoked the recruitment of new, non-fatigued motor units as the explanation for the spectral shift to lower frequencies. Nonetheless, De Luca (1985) stressed the fact that however plausible this explanation could appear, at the time there was no conclusive evidence of the proposed link between spectral modification of the EMG signal and recruitment of additional motor units, a conclusion also reached by Hägg (1992). Therefore, the issue still remains to be clarified beyond doubt.

2.4.1.3 Motor unit synchronisation

The synchronisation of motor units as the muscle fatigues has been also proposed as the mechanism behind the spectral shift to lower frequencies (Lippold et al, 1960; Lloyd, 1971; Chaffin, 1973; Bigland-Ritchie et al, 1981). However, in the view of De Luca (1985) this seems to be the least likely explanation for the spectral modification, because whilst motor unit synchronisation has been reported to appear towards the end of the contraction, the spectral shift is more accentuated at the beginning, and this makes it difficult to establish the one as the cause for the other. Besides, the mathematical models of the electromyographic signal do not support the likelihood of the frequency shift being attributable to motor unit synchronisation (Blinowska et al, 1980; Jones and Lago, 1982).

2.4.1.4 Metabolic factors and muscle fibre type

It has been proposed that the main metabolic factor behind the spectral changes of the EMG signal is the accumulation of acidic by-products, since this means an increase in the concentration of H^+ and a consequent decrease in pH which, in turn, affects the conduction velocity of the action potentials along the muscle fibre (Hermansen and Osnes, 1972; Sahlin et al, 1975; Tesch et al, 1978). It has also been reported that the spectral modifications appear more quickly and are more pronounced in muscles with a high proportion of fast twitch fibres (Komi and Tesch, 1979; Viitasalo and Komi, 1980; Moritani et al, 1982). Alternatively, when the spectral changes in a given muscle have been compared between individuals, those who had the higher proportion of fast twitch fibres exhibited the more pronounced shift to lower frequencies (Viitasalo and Komi, 1978).

2.4.2 Considerations to the use of electromyography during isometric exertion

Although today it appears to be firmly established as one of the most widely used tools in Ergonomics, EMG is far from being a fail-safe technique. There are many factors that render it fraught with the risk of getting the wrong result, most of which lie within the procedures employed in collecting, analysing and interpreting the myoelectric signal.

For the benefit of those without a strong clinical background, but with the need (or the desire) to use EMG, Marras (1987, 1990) identified the most obvious and/or dangerous of such risks. These may be summed up as follows: i) selecting electrodes which are not the most appropriate to get samples from the muscle of interest; ii) if using surface electrodes, the muscle sampling area might change between recordings due to subject's motion, as a result of the muscle's contractile activity, or even if there is a large amount of fatty tissue underneath the muscle; iii) inadequate preparation of the electrodes and/or, if using surface electrodes, the site of application; iv) deficient procedure in attaching or introducing the electrodes; v) poor quality of the signal (most likely due to noise from motion artifacts) which goes undetected; vi) poor conditioning of the signal in the stages of amplification and/or filtering; vii) selecting the wrong analytic approach. Fortunately, besides identifying those risks, Marras also offered advice regarding the precautions necessary to avoid them.

Mirka (1991) highlighted another danger present in the way EMG is sometimes used in Ergonomics, which is highly relevant from the practical viewpoint. He referred to the practice of measuring the EMG activity during a given action of the muscles that operate a joint (whilst the joint is kept at a known angle), and normalising that level of activity against the one registered when the same joint is activated in conditions that will then be taken as reference. He contended that, if a valid normalisation is to be attained, then both the reference action and the one referred to it should be performed in the same conditions, only changing the degree of muscle activation, which is determined by the strength of the action. However, he asserted, it is common practice to normalise the activity registered while the muscles are engaged in more or less dynamic actions (which means that the joint angle is rapidly changing), against that obtained from a single maximal activation of the static joint, which often is not even held at the same angle it was during the referred movement. In Mirka's view, such normalisation procedure will yield grossly inaccurate comparisons.

To prove his point, Mirka registered the activity of the erector spinae muscles during a series of controlled extensions of the trunk (that is, extension at constant force and velocity). He compared that activity against the one observed in two reference conditions: one, the maximal activation of the same muscles at an extension angle arbitrarily chosen, different from those used in the controlled movements; the other, the maximal activation of the muscles precisely at the final angle of the corresponding extension. He reported that the comparison of the two normalised values showed large differences, larger than 75% in some cases, with the value normalised against the arbitrary angle being larger than the obtained from the alternative procedure.

The contributions by Marras (1987, 1990) and the results of the study by Mirka (1991) lead these comments to two conclusions: a) although the technical sophistication of the more up-to-date equipment available has reduced the possibilities of getting it completely wrong, EMG is a technique that requires a lot of attention to detail, and sometimes despite doing so, things could still go wrong; b) the results obtained with this technique need to be carefully assessed and, consequently, interpreted with caution.

2.5 Subjective assessment of fatigue and exhaustion in isometric exertion

The measurement of physiological parameters during static contraction (e.g. heart rate, blood pressure, blood flow through the muscles, metabolite concentration) may help determine with fairly good precision the changes that are symptomatic of fatigue. However, those measurements usually involve procedures that require not only the use of sophisticated arrangements of equipment, but very often the help from a person with a fairly high level of skills in their use, so as to ensure the reliability of the results being obtained, a combination of circumstances that in many cases proves difficult to get. For this reason, assessment methods which make use of the rating by the subjects of their sensations of discomfort, fatigue or pain are especially appealing, since all they require is the use of a scale on which the subject will locate a value attached to a verbal description which they think best reflects their sensations at that moment.

2.5.1 Nature of the relationship between subjective assessment and the duration of the isometric exertion

The collection of subjective ratings has been applied to a variety of situations involving isometric exertion, not necessarily of a postural nature, and in many cases it has produced evidence of a strong linear relationship between the subjective perception and the passage of time. Remarkably, as the studies cited below will show that linear relationship appears to hold irrespective of the nature of the task being performed, the instrument used to collect the subjective ratings, the muscular group most heavily engaged in the exertion, and of the magnitude of the effort involved, .

2.5.1.1 Pain ratings during continuous handgrip

Caldwell and Smith (1966) measured the intensity of the pain provoked by the continuous holding, to the limit of endurance, of a handgrip on a dynamometer, with force corresponding to 25%, 40% and 55% of the subject's maximum capacity for such form of exertion. To test the influence of the restriction of blood flow through the forearm on the intensity of the perceived pain and the endurance to it, the subjects performed the effort with and without the presence of a pneumatic cuff wrapped around the upper arm and inflated to 15 mm Hg above the systolic pressure. The subjects were asked to rate their sensation of pain using a scale marked with the units 1 to 5, the marks corresponding to an intensity of pain described as 'just noticeable', 'moderate', 'severe', 'very severe' and 'intolerable', respectively. The times when the subject let the experimenter know that they had reached the next point on the scale were recorded.

When considered against the actual length of the contraction, the pain increased at a different rate for each force applied, with the higher ratings returned earlier within the exertion at the higher force; also, at a given force level the mechanical impediment of the blood flow accelerated the increase of pain. However, those differences disappeared when the contraction times were expressed as percentage of the endurance to the effort, since the various ratings were returned at the same point within the duration of the exertion, irrespective of the proportion of the strength applied and of the presence of the restriction to blood supply. The plot of the pain intensity against the percentage of the endurance time (shown in figure 2.5a) exhibited a strong linear trend.

2.5.1.2 Pain ratings during load holding

Kirk and Sadoyama (1973) also used a 5-point scale to collect ratings of perceived pain from subjects who were asked to hold a load with their right hand whilst the arm was displaced at an angle of either 10° or 20° to the side of the body; the load was adjusted to represent 30%, 50% or 70% of the heaviest load the subject could handle in each posture. All the efforts were exerted to the limit of endurance and the subjects were asked to rate the pain every 30 seconds.

Figure 2.3 shows the pain ratings obtained at each of the three exertion intensities plotted against the holding time expressed as percentage of the subject's endurance. The points fell on straight lines with very similar slopes which means that, despite the significant difference between the loads being held, the subjects perceived the pain to have reached the same intensity after they had been exerting the force for approximately the same proportion of their maximum endurance. Also, particularly at the highest level of force, the increase of the pain over the holding time followed the same pattern whether the load was held with the arm at an angle of 10° or 20°, which means that the subjects perceived the pain in the same way despite the difference in the biomechanical load imposed on the arm.

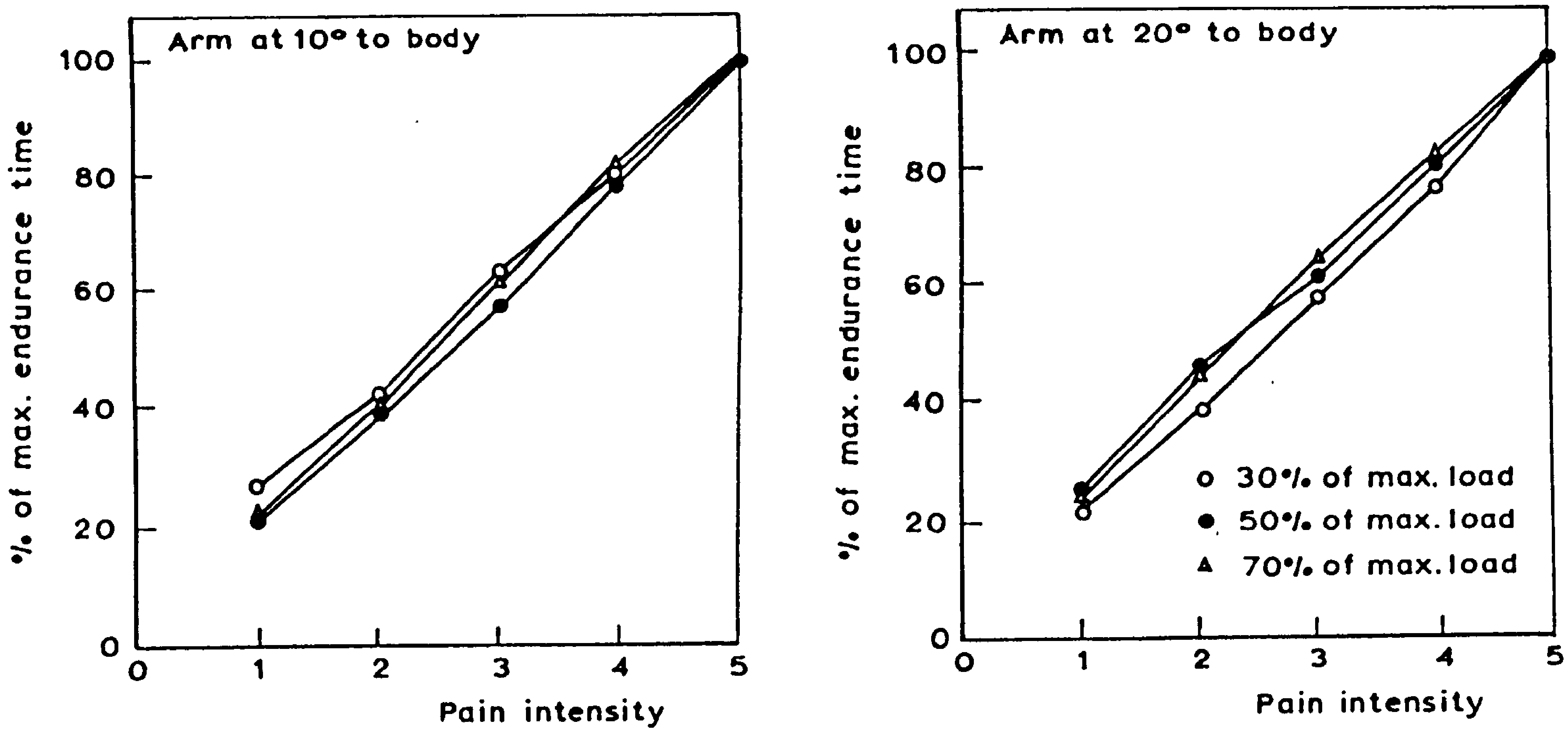


Figure 2.3 Ratings of pain intensity returned during static arm work in two positions. (From Kirk and Sadoyama, 1973).

2.5.1.3 Ratings of discomfort and pain from the passive loading of joints

Harms-Ringdahl et al (1983) studied the increase of discomfort and pain as perceived by subjects who had the joint of either one elbow or one knee moderately loaded whilst kept in an extreme position. They used the 10-point scale developed by Borg (1982), which spans between the complete absence of discomfort (rating 0) and a degree of discomfort such that it makes imperative to end the effort (rating of 'maximal', beyond 10). Given the nature of the stressor being applied, to avoid the risk of causing damage to the subject the experiments were always halted when the subject reached a rating of 7, described as a sensation of very strong discomfort. The results showed that the discomfort grew following a straight linear pattern for both joints tested. When the discomfort ratings were related to the product of the loading moment by time, it emerged that the sensitivity to pain and discomfort in the elbow joint was six times that present in the knee joint.

2.5.1.4 Discomfort ratings whilst enduring postural loads

Manenica (1986) assessed the increase of the discomfort as perceived by subjects who performed a tapping task to the limit of their endurance whilst keeping one of seven postures that involved different extents of trunk flexion and arm extension. The subjects returned ratings of discomfort every 30 seconds, using the 20-point scale proposed by Borg (1973). For each of the seven postures, a regression line was fitted to the average of the discomfort ratings returned by the subjects at the times corresponding to 25%, 50%, 75% and 100% of their maximum endurance. The postures differed significantly in regards of the body parts that experienced the worst discomfort (probably caused by their bearing the heavier biomechanical loading); despite this, the subjects tended to reach similar levels of discomfort after a given proportion of their endurance had elapsed. This made the seven regression lines very similar to each other, as shown in figure 2.4.

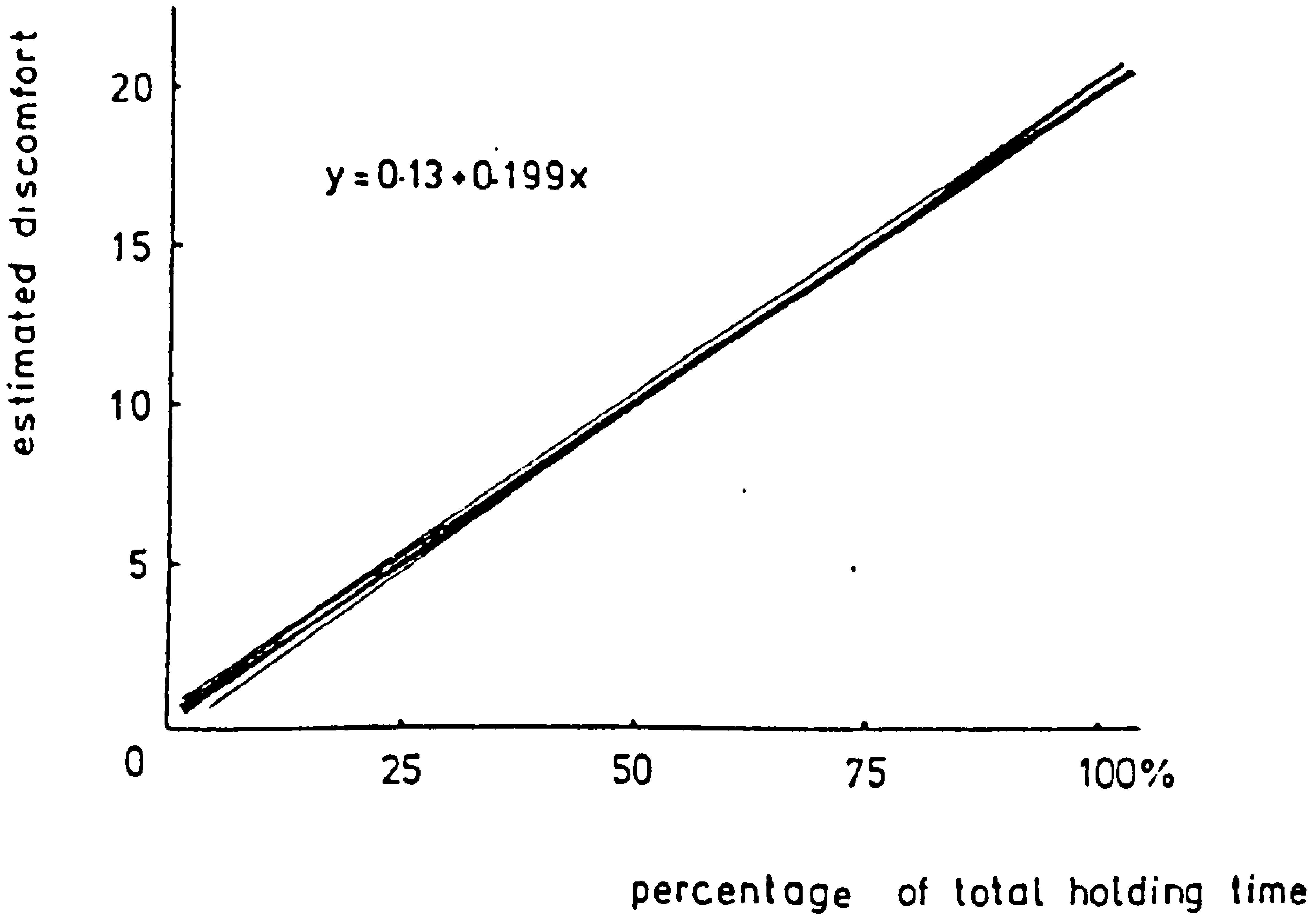
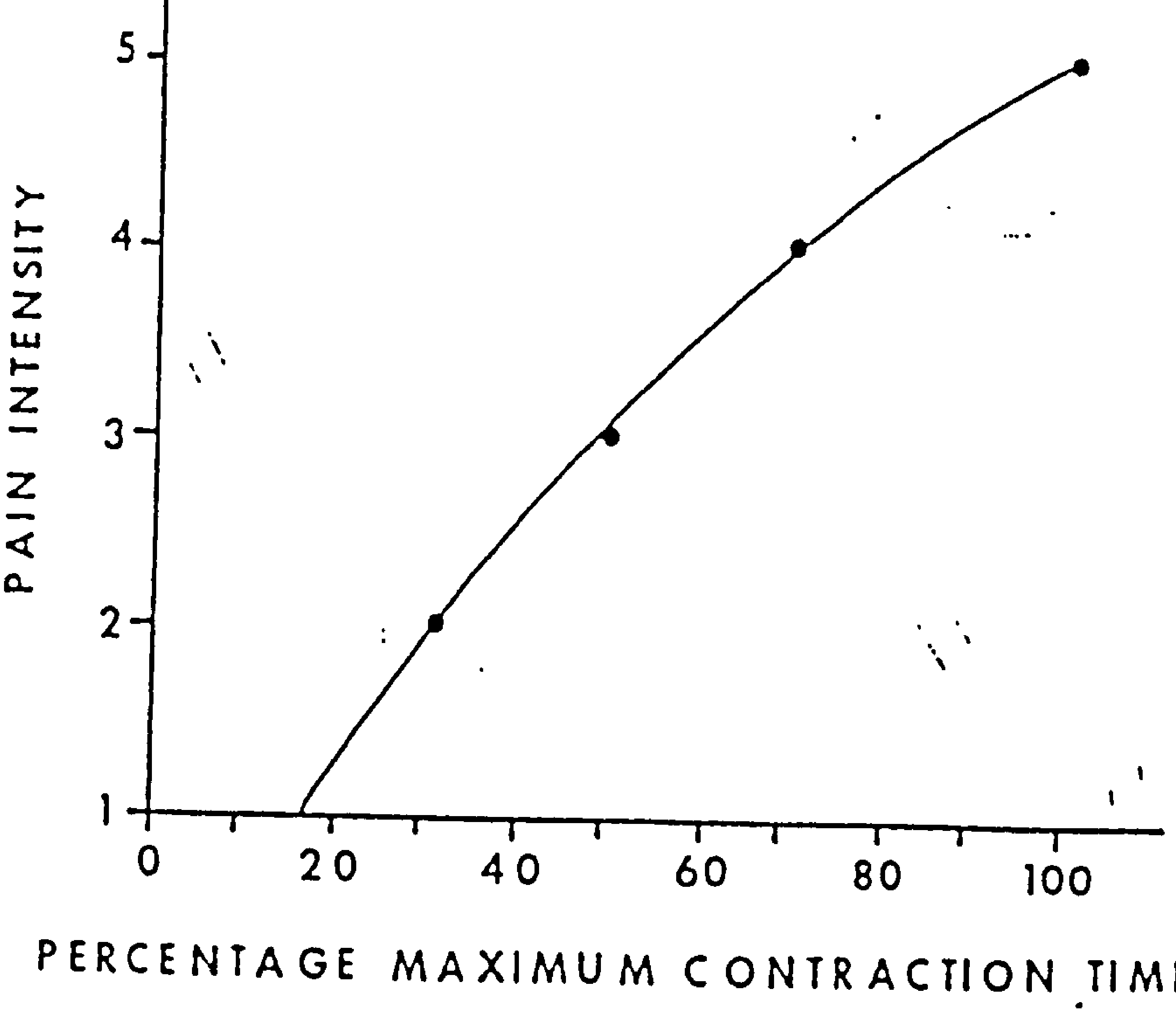


Figure 2.4 Relationship between postural discomfort and the relative holding time during the performance of a tapping task in seven postures with varying degree of trunk flexion and arm extension. The regression equation corresponds to the overall relationship. From Manenica, 1986.

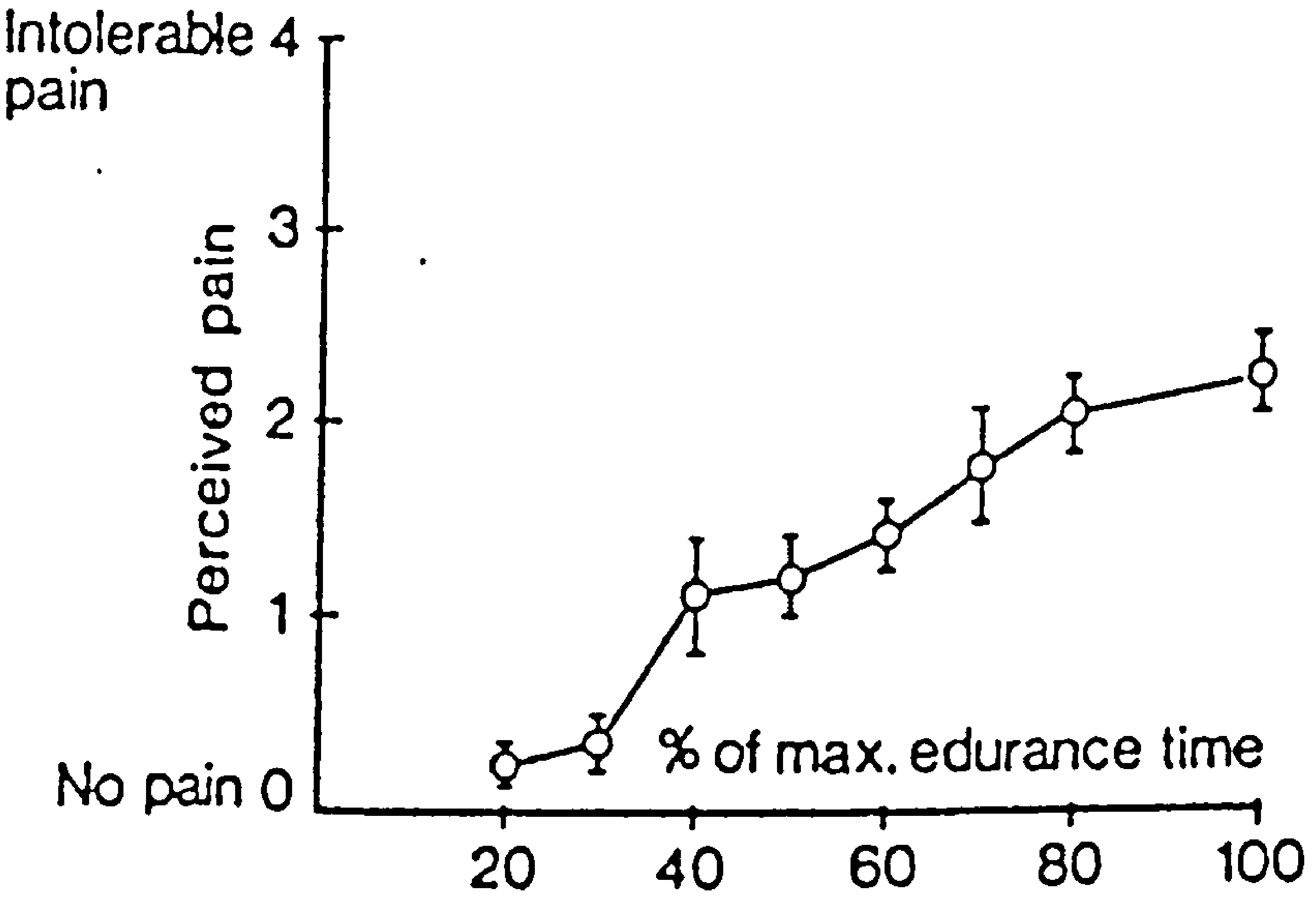
2.5.2 Considerations to the linearity of the relationship between subjective ratings and duration of isometric exertion

Kilbom et al (1983) did not find a linear relationship between perceived pain intensity and the duration of the exertion when 18 subjects performed static flexion of the right elbow with a force equal to 25% of their MVC. The maximal strength and the endurance to the effort applying 25% of that force were measured three times, in the first, second and final sessions; in other sessions the exertion was stopped by the researcher after the subject had sustained it during a time equivalent to 20, 40, 60 and 80% (8 subjects), or 20, 30, 50, 70 and 80% (10 subjects) of the maximum endurance achieved during the second measurement. The pain ratings were collected in a way similar to that applied by Caldwell and Smith (1966). The scale used had 5 points, the lower end (marked 0) corresponding to a state of "no pain" and the higher end (marked 4) was labelled "intolerable pain"; however, instead of reporting on the increase of the perceived pain whilst sustaining the effort, the subjects only had to give a rating immediately after the exertion had finished either through exhaustion or been halted by the experimenter. The ratings returned by the subjects at the end of each session were averaged and then set against the corresponding proportion of the endurance time.

Figure 2.5 illustrates the contrast between the results obtained by Caldwell and Smith (1966) and by Kilbom et al (1983). There are two fundamental differences, which the latter authors very much emphasised. One is evident at the lower end of the rating scale: whilst the subjects in the early study reached the consecutive ratings 2 to 5 at intervals of quite similar duration (roughly 25% of the endurance time), the majority of the subjects in the later study perceived "no pain" at all, even when they had already used up to 30% of their maximal endurance.



a) From Caldwell and Smith, 1966.



b) From Kilbom et al, 1983.

Figure 2.5 Pain ratings returned by subjects in two studies of continuous static contraction. a) Average values returned during the exertion, at 25%, 50%, 75% and 100% of the maximum contraction time; b) mean and SEM of the values returned at the end of contractions lasting between 20% and 100% of the maximum endurance time, in intervals of 10%.

The second (and more remarkable) difference between the results reported by Caldwell and Smith (1966) and by Kilbom et al (1983) appears at the higher end of the scale: whilst every one of Caldwell and Smith's subjects reached the highest level of perceived pain by the end of the exertion, none of Kilbom et al's subjects reported to be experiencing "intolerable pain" even when they had sustained the exertion to their limit of endurance.

Those two differences have been attributed to the presence of a timing artifact in the subject-paced rating procedure applied by Caldwell and Smith (1966) since, according to Kinsman and Weiser (1976), when asked to report at what point had they reached the consecutive marks on the rating scale, what the subjects actually did was either to adjust the rating to the proportion of the endurance they had already used, or simply to replicate the interval it took them to reach the first rating mark. Kinsman and Weiser stated that such procedure is bound to result in a linear relationship between subjective ratings and time elapsed but, they affirmed, the relationship is actually a spurious one.

This issue, however, was addressed in a study by Menzer et al (1969) in which sixteen subjects performed handgrip to exhaustion on a dynamometer, applying 25% or 40% of their maximal strength. The aim of that study was to compare the ratings obtained through the subject-paced method with those generated when they were clued to return a rating at irregular intervals which were determined at random. Besides the variant introduced to the rating procedure, the subjects were also asked to pick their ratings from one of two different scales: one the 5-point scale used in the earlier studies (Caldwell and Smith, 1966), and another one with 10 rating marks. Menzer et al (1969) found strongly linear relationships between the actual exertion time (not proportion of the maximum endurance) and the pain ratings, for all the combinations of force level (25% and 40% MVC), scale (5-point or 10-point) and rating modality (self paced and cued at random).

2.5.3 Issues concerning the application of subjective ratings

Although undoubtedly simpler in their application than most of the techniques based on the evaluation of physiological changes, the use of subjective ratings is also surrounded by a number of complex issues. Whilst some of these issues are of entirely conceptual nature, there are others that relate to the use of the subjective perception as a tool for the assessment of fatigue. The most relevant among both categories will be reviewed now.

2.5.3.1 Conceptual considerations

At the very core of the conceptual matters surrounding the use of subjective ratings lies the difficulty in defining precisely what is the sensation being rated. Pain and discomfort are the terms most frequently used in this context, but it is true that they could mean all sort of things to different people. Certainly, discomfort is generally associated with a sensation of acute fatigue, but it cannot be said that whenever discomfort appears it will lead to an impairment in performance, an occurrence that by itself defines the appearance of what Bills (1943, cited in de Luca 1985) called objective fatigue. Furthermore, Kuorinka (1983) drew attention to the fact that, by and large, the concept of discomfort is even less coherent than the concept of fatigue is. Also, pain may be perceived as just one of the manifestations of discomfort (Kuorinka 1983), and in some instances pain and discomfort have been incorporated into a single entity, as in the study by Harms-Ringdahl et al (1983) mentioned earlier in this chapter.

Another important issue relates to the fact that it is virtually impossible to ensure that the subjects are rating the sensation they are supposed to. Thus, Caldwell and Smith (1966) stressed that they could not be sure about what exactly their subjects were rating. Although the subjects were instructed to concentrate upon the intensity of the pain they were experiencing, they could possibly have been rating their perceived exertion, or even the reserve of strength they judged to have left at the moment of returning a rating. Kuorinka (1981) also raised the question of whether a subject is actually rating perceived exertion when asked to rate discomfort. Thus, when the researcher is asking for the rating of a perceptual phenomenon such as pain, perceived exertion or discomfort, he could be getting information regarding any of them. Furthermore, even though strictly speaking they do not define the same thing, in the everyday usage the terms fatigue, discomfort or pain tend to be used as synonymous.

2.5.3.2 Practical considerations

At a more practical level, the use of subjective ratings raises two very important issues. The first of them is whether it is possible to establish a significant and, above all, reliable relationship between the subjective perception of fatigue and the various physiological criteria which have been taken to indicate the presence and extent of such phenomenon. Naturally, it is hardly surprising to find reports on this matter which differ widely. For example, Kilbom et al (1983) carried out an investigation with the explicit aim of studying the development of fatigue and the relationship between psychological and physiological indices of fatigue. They asked 18 male subjects to hold the elbow of their dominant arm flexed at 90° whilst exerting a force equivalent to 25% of their maximal flexion strength. Each subject's maximum endurance to the exertion (T_{max}) and maximal flexion strength (MVC) were measured three times (T_{max1}, T_{max2}, T_{max3}; MVC_1, MVC_2, MVC_3). In addition to the three exertions to maximal endurance, eight of the subjects (group I) performed exertions that lasted either 20, 40, 60 or 80% of their T_{max2}; the other ten subjects (group II) carried out holdings that lasted for 20, 30, 50, 70 and 80% of their T_{max2}.

In order to obtain the physiological indices of fatigue, during the measurement of T_{max2}, intraarterial blood pressure (through a catheter inserted into the brachial artery on the non-dominant body side) was measured continuously on the subjects of group I, and EMG from the exercising biceps brachii muscle was recorded on all the subjects; heart rate was measured continuously during all the exertions, on all the subjects. To assess the psychological dimension of fatigue, the subjects returned ratings of perceived pain immediately after finishing each exertion, whilst ratings of the perceived effort expenditure were returned only after the sub-maximal exertions. The scale and the method used to obtain those ratings, as well as the procedures applied to evaluate the results have already been described in section 2.5.2.

Kilbom et al (1983) used the value of the centre frequency of the EMG signal as a measure of local muscular fatigue. The mean amplitude of the signal was used as a measure of the activation in the central nervous system. When the changes in these physiological indices were related to those that occurred in the subjective perception, the authors declared themselves surprised by the results. In summary, they found that the relationship between the centre frequency and the perceived effort expenditure was rather poor, as was the relationship between perceived effort and both EMG amplitude and heart rate. The best fit was obtained between perceived effort expenditure and the intraarterial blood pressure.

These findings contrast markedly with those reported by Rohmert et al (1986). They studied the endurance to the loads imposed on muscles and skeletal structures by five working postures. Whilst two of those postures were designed to place the main stress on muscles of the upper limbs, the other three were intended to put the strain on the back/trunk complex. Each posture was tested with a series of external loads that represented proportions of the maximal strength of the subject. During the trials, EMG signals were collected from the muscles known to be the most burdened in each of the postures. In addition, the subjects returned ratings of perceived exertion using the category-ratio scale developed by Borg (1982).

The trials with the two postures designed to put load on the upper limbs yielded the clearer results. These showed that the ratings of perceived exertion coincided with the EMG measurements, and the degree of coincidence was such that Rohmert et al concluded that the subjective assessment might render the measurement and calculation of the load levels unnecessary. This result, it is worth reiterating it, differs widely from those reported by Kilbom et al (1983).

The second practical issue is whether the perception by the subject is in itself a reliable means to assess the extent of the loads created by the static exertion on the muscles and related structures, and to determine when the muscles have reached the point where it is not possible to sustain the effort any longer. Contrasting points of view in this respect may be found in Corlett and Manenica (1980, p 8) who considered that "energy expenditure and postural pain represent, in most cases, independent criterion limits to performance..." and in Kilbom et al (1983, p 191) who reached the conclusion that "perceived pain on the local muscles was not the primary factor limiting endurance on the task".

Thus, it is evident that many matters remain to be settled before it may be said with absolute certitude that fatigue arising from isometric exertion may be precisely defined and measured attending either to the changes in the physiological state of the subject, or to their perception of how fatigued they are, or even to a combination of both criteria.

2.6 Postural loads and their effects

2.6.1 'Good' posture versus working posture

Maintaining the body in a certain posture is the result of isometric contraction; if the posture is maintained for a long time, then that isometric effort becomes a factor in the development of fatigue (Monod, 1972). Roaf (1977) asserted that from a purely physiological point of view, a posture may be deemed 'good' if, when adapted to the circumstances, it may be maintained with the minimal muscular effort. Many occupational settings have evolved in such a way that by removing the need for heavy, dynamic activity, the main physical demands on the worker are of postural nature, although this is a situation that may easily go unrecognised (Corlett and Manenica, 1980). Nonetheless, if those demands stemmed only from 'good' postures as defined above, the body should be able to cope with them so that the worker does not get to feel excessively fatigued as a result of having to work for a long time in a fixed posture.

However, very often that is not the case. Roaf (1977) pointed out that, whilst at work, many people are forced to adopt postures that depart from 'goodness'. Those postures are determined by the need to conform to the equipment or machinery they use to carry out their tasks and, in many cases, those implements have not been designed with the posture of the user in mind. The worker is then forced to adapt to the circumstances as best as they can, and this converts the working posture into a position they adopt mainly because it is appropriate for the performance of the task (Corlett, 1981).

2.6.2 Undesirable effects of inadequate working postures

The first and most obvious ill effect of the departure from the 'good' posture that poses the least physiological demands is the appearance of discomfort (Corlett and Bishop, 1976; Wiker et al, 1989; Genaidy and Karwowsky, 1993). However, as noted by Corlett and Manenica (1980), if the awkward posture is held for long enough, discomfort evolves into pain and this will mark the limit to the maintenance of such posture; if the circumstances allow it, the person will shift the posture so as to relieve the pain, but if the design of the workplace prevents this possibility, then the work will be interrupted for long enough to acquire at least momentary relief, and productivity will be affected (Corlett and Manenica, 1980).

But the impairment of productivity, however relevant it may be for the economic functions of work, should not be seen as the main source of concern when dealing with the ill effects of inadequate working postures; equally, if not more relevant are the long-term effects on the health of the worker (Corlett, 1988). Indeed, as Westgaard (1988) pointed out, when those circumstances are forced on a person for long enough, they may eventually have a harmful effect on their physical well being, particularly through the development of muscle injuries. Numerous studies have investigated the possible link between inadequate working postures and the occurrence of musculoskeletal disorders. Those which appeared the more relevant to the present investigation will be reviewed next.

2.6.3 'Bad' posture as precursor of musculoskeletal symptoms

Van Wely (1970) reported on the possible effects that what he called 'bad' postures would have on the musculoskeletal system of the person who was forced to (or even chose) to work in those postures. The criteria applied in that study to classify a posture as 'bad' were that it: 1) overloaded muscles and tendons, 2) loaded joints in an uneven or unbalanced manner, or 3) involved a static load on the musculature. An important feature of the study was that rather than matching them to already known clinical findings, the analysis of the postures was an effort to predict, based on their characteristics, what parts of the body would be affected when they were adopted for long enough.

The end product of the study by van Wely (1970) was a table (reproduced below as table 2.1) relating those postures deemed the most undesirable, to the body sites where their ill effects were most likely to appear.

Table 2.1 'Bad postures' versus probable sites of symptoms

Bad postures	Probable site of pain or other symptoms
Standing (and particularly a pigeonfooted stance)	Feet, lumbar region
Sitting without lumbar support	Erector spinae muscles
Siting without good footrests of the correct height	Knee, legs and lumbar region
Sitting with elbows rested on a working surface that is too high	Trapezius, rhomboideus and levator scapulae muscles
Upper arm hanging unsupported out of vertical	Shoulders, upper arms
Arms reaching upwards	Shoulders, upper arms
Head bent back	Cervical region
Trunk bent forward, stooping position	Lumbar region; erector spinae muscles
Lifting heavy weights with back bent forward	Lumbar region; erector spinae muscles
Any cramped position	The muscles involved
Maintenance of any joint in its extreme position	The joint involved

(Reproduced from van Wely, 1970)

The posture described by van Wely as "upper arm hanging unsupported out of vertical" which was related to the appearance of pain (or other symptoms) on the shoulders or upper arms is highly relevant to this investigation, since it is reasonable to assume that such a heading comprised all forms of arm deviation, including abduction, which was the focus of interest in the present research.

2.6.4 Relation between workplace layout, postural loads and musculoskeletal disorders

Aarås (1987) reported on a long-term investigation of the effects on the musculoskeletal system of postural loads attributable to the layout of a number of workplaces. The study was conducted in a plant manufacturing telephonic equipment, and was prompted by the combination, in a single year, of a high rate of sick leave, an increased number of cases requiring rehabilitation treatment, and a large proportion of labour turnover.

The study involved the redesign of five workplaces in a way that was expected to reduce the size of the postural loads imposed on the worker. Due to the nature of the operations involved in the five workplaces studied, the postural loads were estimated by measuring the electromyographic activity in the descending part of the trapezius muscle on the worker's dominant side, and converting it into a percentage of the activity elicited during the maximum voluntary exertion of the muscle.

The study had three aims: 1) to establish whether there was a quantitative connection between the extent of postural loads and the development of musculoskeletal illness; 2) to evaluate whether the modification of the workplaces did in fact reduce the extent of the postural loads; 3) to assess the influence of the reduction of the postural loads on the incidence of musculoskeletal illness.

Aarås (1987) based his report on data collected during the 7 years following the introduction of the changes; when possible, these were compared against a similar period prior to the changes. He found a definite relationship between the extent of the postural loads and the incidence of sick leave in the period preceding the changes to the workplaces. This was demonstrated by the significantly higher loss of working time through sick leave, for the five workplaces as a whole, when it was compared to the same statistic for a control group formed by workers in the same company who were engaged in general office work. The strong link between postural load and incidence of sick leave was also manifested when the workplaces were compared to each other.

Comparisons between the levels of muscular load before and after the introduction of the changes were only possible for two work situations, one of them involved mostly static activity, the other one was of a more dynamic nature. The reduction of the postural loads produced by the redesign of the workplaces was significantly larger for the more dynamic operation than for the mainly static one. This was interpreted by Aarås (1987) as evidencing that the postural loads of a static nature were less amenable to improvement through the changes implemented. In fact, although the average level of muscular load present in the static operation was nearly halved (from 20% MVC to 11% MVC), this value was still too high compared with the 2-5% recommended by Bjøkrsten and Jonsson (1977). Nevertheless, the changes did achieve their main goal, since even for that workplace the incidence of sick leave due to musculoskeletal symptoms was in fact reduced.

2.6.5 Absence of a link between postural loads and musculoskeletal disorders

The strong link between the level of muscular load and the development of musculoskeletal complaint, suggested in the results reported by Aarås (1987), was not seen by Westgaard et al (1991), who applied the same approach in their study of 30 female workers who operated chocolate packing machines. They were first evaluated when they took up the job, to ensure they were not suffering from any severe form of musculoskeletal trouble at that point, and any previous episodes were recorded. The level of load (measured from the EMG activity) imposed by the work on the trapezius muscle was also determined. The whole evaluation was then repeated at intervals of 10 weeks, and the subjects reported whether, since the previous interview, they had been in need of medical advice because of musculoskeletal symptoms; these periodic evaluations were repeated up to the 60th week of employment.

None of the subjects had been affected by the time of the first periodic evaluation (week 10), but after that, a total of 17 workers had to seek medical attention because of some form of musculoskeletal complaint. However, there was not a significant difference between the levels of muscular load for the subjects affected and those for the non-affected; in fact, whilst the load levels for the latter group remained fairly constant between the 10-weekly evaluations, the load for the affected subjects tended to decrease. Thus, Westgaard et al (1991) concluded that, at least for the situation they studied, there does not seem to be a link between occupational muscular load and the development of musculoskeletal complaint. They suggested that it is not the load in itself, but the sensitivity of the subject to the load which determines the appearance of the trouble.

2.7 Main issues in work-related musculoskeletal disorders

Because of their impact on the well-being of the working population and their economic consequences in terms of decreased productivity and rising costs of rehabilitation or compensation, work-related musculoskeletal disorders have been intensively studied, particularly over the last 30 years. Obviously, the effort has been aimed primarily at the identification of those factors present in the work itself and in the work environment that could be linked to the development of the problems. This section reviews the findings from two contrasting but at the same time complementary kinds of studies. It opens with the presentation of the most relevant among the conclusions reached by two long-ranging review studies which considered the significance of a large number of factors in the generation of work-related musculoskeletal complaints. Once this is accomplished, the review will move into the findings of a number of field studies that have looked into the influence of some ergonomic factors on the presence of occupational musculoskeletal disorders of the upper body.

2.7.1 Factors associated with disorders of the neck and upper limbs

Wallace and Buckle (1987) offered a comprehensive review of the most relevant issues addressed by a large number of studies that have dealt with musculoskeletal disorders affecting the neck and the upper limbs. They started by considering the problem posed by the lack of a unified nomenclature for those disorders which have been attributed to the action of work-related factors. After describing some of the most widely used terms they settled for the term "regional musculoskeletal disorders", agreed during the conference on Epidemiology, Rheumatism and Industrial Labour (Hamburg, June 1985). Defining when a person is affected by a work-related disorder has been another area of disagreement between studies; while some researchers will only consider those cases detected through a medical inspection, others will include any absence from work based on a diagnosis of musculoskeletal disorder, and yet another criterion will be the self report of discomfort and pain. Wallace and Buckle attributed the considerable differences in the incidence rate of work-related musculoskeletal disorders reported in several studies to such lack of uniformity in the nomenclature and diagnostic approach, since the rate of incidence very much depends on what criteria have been applied to classify the effects and the probable causes of the disorder observed.

Next, Wallace and Buckle pointed out that musculoskeletal disorders originate from a combination of factors, some of which could certainly be work-related, but major difficulties lie both in identifying the evidence for a causal relationship between occupation and ill-health, and in determining in what proportions the work factors contribute to the existence of the disorder. Indeed, they contended that the efforts in trying to establish the existence of a causal relationship have been hampered by the lack of theories linking specific health effects to specific causal factors.. Therefore, in the absence of clear-cut cause/effect relationships, much effort has gone into establishing the influence of associated risk factors; among the most widely investigated are the type and design of the job, the design of the workstation, the postures this imposes, the use of repeated movements and forces, subject-related variables (e.g. age, gender, health status, work technique), the way the work is organised, and a number of psychosocial factors.

Regarding the possible relationship between work posture and musculoskeletal disorder, Wallace and Buckle mentioned that a constrained posture involves long-lasting static constriction of the muscles with reduction of blood irrigation, and this provokes local fatigue with the consequent symptoms of tiredness, pain and cramp. Therefore, the studies in this area have usually sought a link between prolonged constrained posture and complaints of discomfort or pain in hands, arms, shoulders and neck. Most of those studies have highlighted the importance of movement in connection with conditions of postural constraints, so that when comparing jobs where workplaces and postures are similar but one of the jobs permits more movement and some change in positions, usually a lower percentage of the workers in that job will report painful symptoms. Besides, Wallace and Buckle noted that even though a large number of studies in this area have been concerned with work that by its nature demands a high degree of postural fixity, this has not always been considered as a risk factor on its own.

However, Wallace and Buckle (1987) emphasised that postural constraints may certainly have an additive or interactive role with other factors. Thus, when postural constraints have been considered in combination with aspects of movement and force, the findings point to a higher incidence of musculoskeletal disorder associated with jobs where the work is of a heavy, monotonous nature. Work content and psychological factors have also been found to interact with postural constraints in provoking musculoskeletal disorders; in general, it has been established that factors associated with stress at work (job demands, lack of autonomy, relationship between employer and employee) will determine the presence of muscular tensions that could induce injury. Wallace and Buckle also noted that studies relating psychosocial aspects of work with the presence of musculoskeletal complaints have been conducted mainly in connection with jobs that involve the use of VDTs, and they suggested the need to extend this approach to other types of jobs, particularly those where the tasks incorporate a high degree of automation.

The individual characteristics form the last group of factors that have been assessed for their interaction with posture as the cause for musculoskeletal trouble. Wallace and Buckle (1987) cited the findings of several studies which indicate that body build seems to hardly bear any relationship with the risk of musculoskeletal disorders, whilst mixed findings have been reported with respect to age, since this issue is easily obscured by the fact that older people tend to be more susceptible to arthritic conditions. Finally, Wallace and Buckle stated that the largest individual influence on the likelihood of developing musculoskeletal troubles might stem from the differences in work technique, but this was an area that at the time of their writing appeared to have been neglected in most of the studies.

2.7.2 Relationship between occupational factors and disorders of the soft tissues in the shoulder

Sommerich et al (1993) reviewed the most recent studies referring to the association between occupational factors and the development of soft tissue disorders at the shoulder. This review reached basically the same conclusions as those of Wallace and Buckle (1987), showing that 6 years on the situation had hardly changed. However, by concentrating their attention on studies that had dealt specifically with conditions that affect the shoulder, Sommerich et al were in a position that allowed them to identify the most important factors that have hampered the advance towards a better understanding of the problem, and to suggest ways of avoiding them in future. Thus, Sommerich et al considered that the main problem lies in the lack of uniformity in the identification of cases and in the description of the risk factors. They attributed this problem to three main causes:

a) the existing disease classification systems do not recognise the occupational origin of many musculoskeletal disorders;

b) symptoms are not always easily recognised and do vary in relation to the work the person is performing, and so they are seen to appear and disappear;

c) the devices applied to the measurement of the problem are not always adequate for a thorough quantitative analysis.

In the view of Sommerich et al (1993), the way forward is the establishment of standard methods that may be applied to both the case definition and the measurement of the physical exposure, so that in future the studies should yield information that may be useful beyond the limits of the single plant or department where the study was carried out. What those authors consider the ideal situation is the establishment of longitudinal surveillance studies in which the employees would be screened for signs of exposure at the time of employment, and then monitored periodically. In this way, if a musculoskeletal disorder eventually appeared, it would be possible to determine the dose-effect relationship between the exposure and the magnitude of the symptoms. But the authors also stressed that, despite being incomplete, at the present time there is enough information to allow the modification of workplaces and work practices so as to reduce the impact of musculoskeletal troubles among the working population.

2.7.3 Ergonomic factors associated with musculoskeletal disorders of the upper body

What follows is a brief review of the results presented by a number of field studies whose authors looked in some detail at the ergonomic factors most likely to be associated with the presence of musculoskeletal troubles that affect the upper body, and especially the area of the neck and shoulder.

2.7.3.1 Work mechanisation and static loading of the muscles

Maeda (1977) presented the results of a survey among Japanese workers in a variety of occupations that appeared to be associated with a higher than usual incidence rate of musculoskeletal disorders of the neck and shoulder. Those disorders were grouped under the term 'occupational cervicobrachial disorder' (OCD). As for the causes of the problem, he first and foremost drew attention to the effects that increasing mechanisation had on the tasks performed by workers, for whilst this meant the reduction of heavy muscular exertion, it also increased the localised use of the musculature of the upper extremities. Such change was the most noticeable among people working on assembly lines who, at the time of the survey (1974) had an incidence of 20.9% of work-related excessive fatigue at the shoulder, arms and hands; higher than for any other group of workers.

However, the analysis of the tasks involved in a number of jobs which had been observed to provoke an elevated incidence of OCD revealed that, irrespective of whether or not a machine was involved, the common factor in all the cases was the presence of static loads on postural muscles of the neck and shoulder, as well as static and/or dynamic load on arm and hand muscles. Nevertheless, Maeda (1977) also emphasised that the appearance of OCD could not be attributed entirely to those static and dynamic loads, because it was observed that in many cases the way in which the work was organised provoked a high level of mental strain, and this appeared to be another factor that favoured the appearance of the musculoskeletal disorders.

2.7.3.2 Physical demands of the task, extreme postures and individual characteristics of the worker

Bjelle et al (1979) approached the problem from a different perspective. They studied a group of 20 men who had been suffering from long-term (more than 3 months) shoulder pain believed to be associated with their work, and compared their anthropometric data and several characteristics of their jobs with those of a group of 34 men who performed the same jobs, but were free of any symptom. When compared with this group of referents, the affected workers were significantly older, and the maximum force they could apply in a handgrip evaluation (right hand) was significantly lower, but their stature and body weight were not significantly different. Both groups of workers were also compared with respect to the physical load (light, heavy or very heavy) of the jobs they performed at the time of the study and the one immediately before that; these comparisons yielded non-significant differences. More than half of the people in both groups had been in a very heavy job before, and more than half were now in light or heavy jobs.

The current jobs of both groups of workers were also compared with respect to the load they imposed on the shoulders, which was assessed as a function of whether the hands were placed above shoulder height, and the frequency of such events. The groups differed significantly in this respect; the affected workers spent more time than the referents with their hands above shoulder height, and this was deemed a causative factor for the presence of chronic shoulder pain, although their being older was considered a predisposing factor. The 20 affected workers were evaluated again two years after the study was conducted; eight of them were still employed in the same or in a less heavy type of job, but seven still complained about their shoulders; three were re-training for transfer to lighter jobs, four were on sick leave (although the authors did not specify whether they had been back to work since the time of the study), and five had been granted disability pensions.

The relationship between the adoption of extreme postures and the development of musculoskeletal injuries was also found by Keyserling et al (1987) who studied two groups of workers in an automobile assembly plant. One of the groups was constituted by workers who for the first time had reported the existence of a trunk or shoulder injury that was provoking persistent pain; the other group was formed by workers who had neither sought medical attention nor were affected by persistent pain. Injured workers and referents were matched attending to the tasks they performed, and these were analysed to determine how often and for how long the workers had to place their trunk and/or arms in postures that deviated from the neutral position. Trunk deviation was incurred when the worker, either standing or seated, had to twist or bend forwards, backwards or sideways by more than 20°. If the forward bending was more than 45° it was called hyperflexion and classified as a separate posture on its own. Equally, the shoulder was considered deviated from the neutral position when it was flexed or abducted by more than 45°, and if the angle was more than 90° it would constitute hyperflexion or abduction.

When the two groups were compared, it emerged that the injured workers were roughly five times more likely to work with the trunk in mild flexion (45 to 90°) for any length of time than the referents, and severe flexion (>90°) or a combination of bending and twisting sideways were six times more likely among the injured workers. Regarding the shoulder, the injured workers were two to three times more likely to work with at least one shoulder in severe flexion (>90°), and they also worked longer with each shoulder in that posture. But besides demonstrating that awkward postures may be a contributing factor to the development of musculoskeletal disorders, Keyserling et al (1987) also pointed out that in the last instance those postures are the end result of the interaction of several ergonomic factors, which include unsatisfactory workstation layout, tools or equipment not properly selected or designed, incorrect work methods and the anthropometric characteristics of the worker.

Schierhout et al (1993) assessed the influence of working posture, repetitiveness of movements and forcefulness of exertion present in the workplace, on the incidence of self-reported musculoskeletal pain. They studied a variety of workposts in industrial plants representing a wide spectrum of processes, but all the posts imposed large physical demands on the worker. The main factors were quantified by assigning a score on a scale of 1 to 4, depending on the proportion of the working time spent in a certain posture or exerting a certain force, and the frequency of repetition of a movement. Schierhout et al found that the adoption of unnatural postures was significantly associated with musculoskeletal pain at any site in the body. The single most frequent unnatural posture among the workers was the holding of the unsupported arms between elbow and shoulder height, and this factor was significantly associated with the incidence of musculoskeletal pain in the neck and shoulders area.

2.8 Arm abduction as an ergonomic stressor

Sommerich et al (1993) wrote that epidemiological studies have produced ample evidence pointing to certain occupational factors which appear to be consistently linked to the presence of chronic shoulder disorders among specific groups of workers. The most prominent among those occupational factors (not listed in order of relative importance) are heavy lifting, highly repetitive motions and the prevalence of awkward static working postures. Arm abduction is one of those awkward working postures. A number of studies, both in the laboratory and on the field have identified it as an important stressor that quickly provokes discomfort and fatigue and that, if allowed to act for long enough, may lead to harmful effects on the anatomical structures in the shoulder region. This section reviews some of those studies and their main findings.

2.8.1 Electromyographic evidence of muscular fatigue provoked by arm abduction

Chaffin (1973) observed the course of the subjective responses of 5 male subjects, and the electromyographic changes in their medial deltoids when they were asked to hold both arms abducted at angles of 30, 60, 90 and 120°, recording the time it took them to reach what he called 'class II' muscular fatigue, described as "cramping continuous with deep hot pains intermittent". Chaffin found that the fatigue rate increased exponentially with the increase in the abduction angle. Whilst at 30° the average time in which the subjects reached class II fatigue was about 68 minutes, at 60° it was 25 minutes, at 90° it was 10 minutes, and at 120° it was 8 minutes.

Herberts et al (1980) found that shoulder abduction, when combined with shoulder flexion, contributed importantly to the development of localised fatigue in some of the muscles around the glenohumeral joint. They collected intramuscular EMG signal from anterior and medial deltoids, supraspinatus, infraspinatus and trapezius muscles on the right side, whilst the subject held a 2 kg weight in their right hand and placed it in 8 different positions, defined as a combination of flexion and abduction of the shoulder, with the elbow always flexed at 90°. Changes in the rate of fatigue development for the anterior deltoid occurred when, with the arm flexed at 90°, the subject increased abduction from 0 to 45° and from 0 to 90°; the supraspinatus showed increased fatiguability when with the arm flexed at 90°, the abduction increased from 0 to 45° and from 45° to 90°; the trapezius muscle fatigued more quickly when the abduction increased from 45° to 90°, with the arm flexed at 45°.

Hagberg (1981a) studied the changes in the electromyographic signals collected from trapezius, supraspinatus, infraspinatus, medial and anterior deltoid, and biceps brachii muscles, that occurred whilst female subjects held the right arm abducted at 90°. He found that all the muscles developed electromyographic signs of fatigue, with increase of the amplitude and decrease of the mean power frequency. The quickest to show increase of amplitude was the supraspinatus muscle (average of 16.2 seconds from the start of the effort) and the slowest was the medial deltoid (124.9 seconds on average). Decrease of the mean power frequency was first shown by the anterior deltoid (23 seconds on average) and last by the infraspinatus (96 seconds).

2.8.2 Arm abduction as the cause of localised discomfort

Genaidy and Karwowski (1993) collected the ratings of discomfort generated when laboratory subjects were asked to perform, one at a time, several of the possible movements of the body segments linked by the major joints, and to hold for 30 seconds the extreme posture resulting from each movement . The subjects returned discomfort ratings taken from a scale marked 0 to 10 in unitary increases. The scale had three anchor points: 0= no discomfort, 5= moderate discomfort, 10= extreme discomfort. The ratings were then averaged over the number of subjects, and a ranking was accorded to each form of deviation around the corresponding joint, the highest ranking going to the posture that generated the highest average discomfort rating.

The movements of the arm around the shoulder joint investigated by Genaidy and Karwowski (average discomfort rating between brackets) were: forwards flexion (4.8), backwards extension (5.5), abduction (4.9), adduction (4.6), medial rotation (4.8) and lateral rotation (4.9). It is remarkable that all the forms of arm deviation around the shoulder provoked feelings of moderate (or slightly higher) discomfort in just 30 seconds of holding. Even though abduction was the second most stressful form of deviation, it has to be considered that extreme backwards extension, which generated the highest discomfort rating, is a posture adopted less frequently both in everyday life and in occupational activities.

2.8.3 Arm abduction as a factor in a variety of work-related illness

Besides the ample evidence gathered from laboratory studies, arm abduction has also been associated with the development of a number of work-related morbid conditions set in the shoulder area.

For example, Sällström and Schmidt (1984) compared the prevalence of thoracic outlet syndrome, a condition provoked by the continued compression of the nerves and blood vessels located between the neck and the shoulder (Putz-Anderson, 1988), among groups of cash register operators, heavy industry workers and office staff engaged mainly in word processing tasks. The condition was significantly more frequent among the heavy industry workers and cash register operators. This was attributed partly to their having to adopt awkward postures, including considerable degrees of arm abduction, that for the cash register operators could be beyond 45°.

Maeda (1977) mentioned the presence of considerable arm abduction among sewing machine operators (left arm) and among amplifier assemblers (right arm). He considered that the abduction of the arm provoked increased static load on the muscles of the shoulder, and this in turn favoured the presence of musculoskeletal disorders in shoulder and arm.

Kilbom et al (1986) carried out a 2-year long study on the female workers of an electronics assembly plant. They started by assessing the strength of the relationship between postures and movements and the presence of musculoskeletal disorders. At this stage, arm abduction in the range 0-30° was identified as a risk factor for the incidence of trouble in the neck and trapezius area. One year later, arm abduction higher than 30° was related to an increase of the problems in the same areas, and two years on from the initial assessment, the abduction of the arm was again an important factor, although by this time the effect was seen as pain associated with all forms of shoulder movement.

However, there have been cases where the relationship between arm abduction and the appearance of musculoskeletal trouble affecting the upper arm and shoulder regions has not been evident. An example is the study by Fine et al (1986) who, assuming that arm abduction is an important contributor to the appearance of supraspinatus tendinitis (as suggested by Hagberg, 1982), compared the length of time spent by two groups of workers with their arms in abduction or flexion beyond 60°. Whilst the members of the first group were engaged in jobs which had been associated with high prevalence of supraspinatus tendinitis, the second group of workers were engaged in jobs that were less likely to provoke such disorder.

Fine et al found that, on average, no group spent more time than the other in the postures of interest; rather, the main difference between the two groups of jobs appeared to be the level of force they involved. Furthermore, Fine et al (1986) suggested that the external rotation of the arm is another postural factor that may have a role in the pathogenesis of disorders affecting shoulder and upper arm, a suggestion also put forward by Owen (1969).

2.9 Physiological mechanisms involved in the musculoskeletal disorders affecting the shoulder

Hagberg (1982) reviewed the physiological mechanisms most likely to be involved in the development of muscular disorders that arise in response to local strain on the shoulder. He classified those reactions in terms of how quickly they occurred in relation to the presence of the strain, and for how long their effects were evident. According to these criteria, the reactions may fall under the headings 'immediate', 'delayed' or 'chronic'. The more important among the reactions belonging to each of these three groupings are described in this section.

2.9.1 Immediate reactions to muscular strain on the shoulder

Hagberg (1982) called 'immediate reactions' those symptoms and disorders that occur immediately during or after exposure to local muscular strain. These may take the form of a mechanical failure, like the rupture of muscles or tendons that occur when the shoulder is suddenly overloaded; fortunately, this is a relatively rare event in most work situations. Discomfort and eventual pain are by far the most common form of immediate reaction to local muscular strain; however, this reaction may not originate only in the muscle itself, since tendons, joint capsules and ligaments can also contribute to the unpleasant sensations. Ischaemic effects are a third form of immediate reaction. These effects appear because the increased contraction levels cause an increase of the intramuscular pressure, which in turn creates an impairment to the blood circulation of the muscle and the clearance of metabolites and, if this condition lasts for long enough, the pH will fall to a level that interferes with the normal function of the muscle's enzymes, contributing to a reduction in strength, co-ordination and endurance.

2.9.2 Delayed symptoms in response to heavy shoulder usage

Muscle soreness that appears between 1 and 3 days after performing unaccustomed occupational tasks was placed by Hagberg (1982) under the heading of 'delayed symptoms and disorders'. Four mechanisms have been proposed as the possible cause of muscle soreness. One of them is the rupture of myofibrils, but the author considered that this should occur only when the subject had performed extremely heavy work involving eccentric contractions. Exudative peritendinitis is the second possible mechanism, and it might be induced by the performance of highly repetitive contractions. Lesions caused by ischaemia have been proposed as a third mechanism behind muscle soreness, particularly for isometric contractions, but it is very difficult to prove the existence of a cause-effect relationship, because the ischaemic effects disappear as soon as the contraction ceases. However, ischaemia in the shoulder muscles may develop very quickly when working with the arms elevated and, if that posture occurs frequently enough, it may lead to cumulative ischaemic trauma (Bjelle et al, 1981). Energy depletion is the fourth mechanism proposed as an explanation for the appearance of delayed muscle soreness, since it has been seen that when the intramuscular demands of energy exceed the metabolic production the result seems to be pain. Hagberg et al (1982) have suggested that the extent of the metabolic stress on the muscles is reflected by the serum level of creatinkinase, because the efflux of this enzyme from the muscle tissue is dependent upon energy depletion. They found increased serum levels of creatinkinase in groups of welders, assemblers and cash register operators; in the latter case they attributed the increase to the constant presence of a low-level static muscular load, a consequence of their having to work most of the time in a fixed position.

2.9.3 Chronic symptoms as a consequence of long-term repetitive usage

When the shoulder muscles are subjected to repeated local muscular strain for a long time, the result will be the development of chronic symptoms and disorders (Hagberg, 1982). Degenerative tendinitis that affects the supraspinatus tendon has been linked to prolonged and repeated work with arms elevated; the likely mechanism is through cumulative ischaemic trauma that induces cellular degeneration, chalk deposits, and ultimately tendinitis. Reactive tendinitis/myalgia develops as a reaction to acute infections; Bjelle et al (1981) hypothesised that the constant presence of local muscular strain on the shoulder region might localise the onset of such reactions to that area. Chronic myalgia is the most typical long-term reaction to local muscular strain. It has been found among workers in whom the usual laboratory tests failed to prove the existence of a rheumatoid disorder (Bjelle et al, 1979, 1981). Hagberg (1982) suggested that this form of chronic pain is due to the existence of a vicious circle, which is illustrated in figure 2.6.

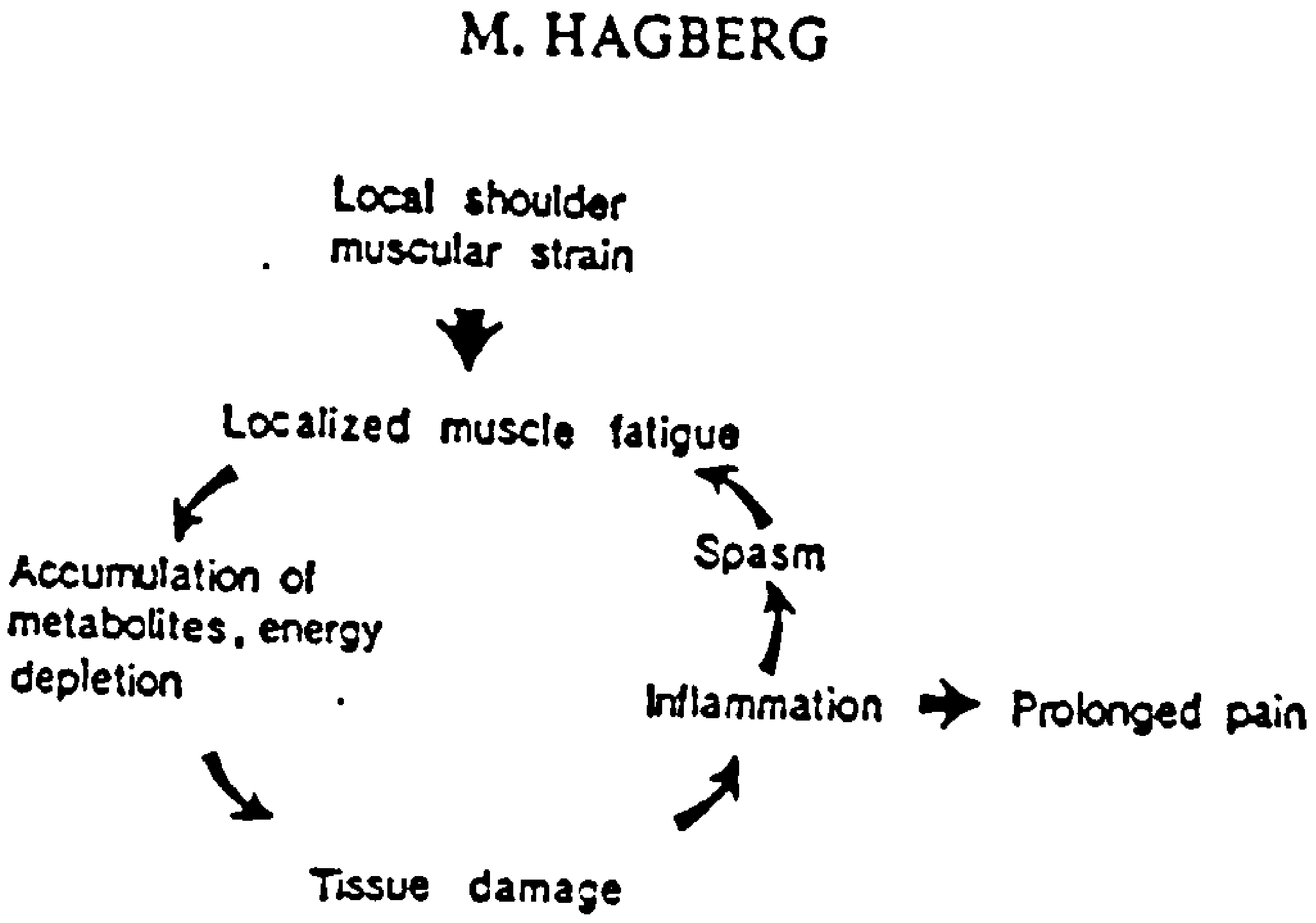


Figure 2.6 The repeated exertion of the shoulder muscles initiates a chain of reactions which induce tissue damage and inflammation that result in continuous contraction (spasm) of the muscle fibres surrounding the damaged area. The spasm itself may cause muscular strain, and a vicious circle producing prolonged pain develops. (From Hagberg, 1982).

Elaborating further on the role of occupational stress as a causal factor for the development of chronic musculoskeletal disorders of the neck and shoulder, Hagberg (1984) expressed two all-important considerations. The first one refers to the possibility that future advances in the knowledge about the aetiology of rheumatic and neuromuscular diseases might prove that long-term muscular pains and disorders which today are attributed to the occupation of the person are in fact well defined forms of illness as yet not clearly understood. The second one refers to the dose-response relationship between musculoskeletal stress and the development of a morbid condition. Hagberg (1984) emphasised that such is an area still not extensively investigated, although it is essential to establish whether it is possible to set limits of exposure that could prevent damage to muscular and related structures.

2.10 Models of fatigue development and recovery in isometric exertion

Establishing a model for the development of fatigue that results from the continuous exertion of a force, and consequently setting a time limit to the effort before undue fatigue arises has been one of the most sought-after goals in Ergonomics. Working towards that end, Monod (1956) and Rohmert (1960) coincided in two basic conclusions: 1) there is an inverse relationship between the amount of force employed in an isometric exertion and the length of time this may be sustained before fatigue reaches a level where the exertion has to be stopped; 2) if the force does not exceed 15% of the maximal strength of the muscle considered (maximal voluntary contraction, or MVC), the exertion may be sustained for a very long time.

However, despite such important similarities, the models for the calculation of the temporal limit for isometric exertion generated by those two seminal investigations were not equivalent to each other, as Drury and Spitz (1978) pointed out. Those models will be reviewed first in this section, followed by the review of a recently proposed model (Deeb et al, 1992) which was developed applying a methodology whose authors (Deeb and Drury, 1990) claim it to be a much better approach than those taken by the authors of earlier works. The model developed by Rohmert (1973) to calculate the length of rest necessary to achieve recovery following the exertion of a constant isometric force for a known time will be reviewed next. The section concludes by reviewing the experimental evidence that has been generated by a number of studies which have put to the test two key assertions put forward both by Monod (1956) and Rohmert (1960): a) that an isometric exertion which involves a force below 15% MVC is practically non-fatiguing and may be sustained for very long time, and b) that such a 'cut-off' force is the same for any muscular group in the body, and for any person.

2.10.1 Monod's model for the calculation of the limit time

Monod (1956) measured the maximum length of time (the limit time) that four different muscle groups could maintain isometric exertions of known strength; he developed the generic expression:

$$t = \frac{k}{\left[\frac{F}{F_{max}} - \frac{f}{F_{max}}\right]^n}$$

in which t is the limit time (minutes), k is a constant, F is the strength applied, F_{max} is the maximal strength of the muscle, f is the strength at which the contraction time tends towards the infinite, and n is an index related to the mechanisms through which the exerted strength influences the limit time. Commenting on this model, Monod (1972) formulated two particularly interesting contentions: a) 'f' (which he called critical strength) is the delimiting value between exertions that will provoke exhaustion and those that may be sustained without fatigue for a very long time; he placed this value between 15% and 20% of the maximum strength of the muscle; b) as long as the strength being applied (F) is expressed as a proportion of F_{max}, the formula to calculate the limit time will be valid for all muscles.

Monod and Scherrer (1965) presented Monod's formula as:

$$T_{max} = \frac{2.5}{(P - 0.14)^{2.4}}$$

in which P is the force applied, expressed as a proportion of the maximal strength of the muscle and 0.14 is the value of the critical strength; in other words, they considered that when the force applied is less than 15% of the maximum strength of the muscle, the exertion may be maintained practically ad infinitum. These authors also suggested that, once T_{max} is known, the strain on the cardiovascular system will be minimised if contraction forces between 15 and 40% of F_{max} are sustained for no longer than two thirds of T_{max}, and forces above 40% F_{max} should be sustained for no longer than one third of T_{max}.

2.10.2 Rohmert's model for the calculation of the maximum exertion time

Rohmert (1960, 1965) also studied the relationship between the strength applied in an isometric exertion and the longest time this can be maintained; he observed the behaviour of 13 muscle groups, on a sample of 25 male and 18 female subjects. He found that the relationship was expressed by the formula

$$T_{max} = -1.5 + \left[\frac{2.1}{F}\right] - \left[\frac{0.6}{F^2}\right] + \left[\frac{0.1}{F^3}\right]$$

in which T_{max} is the time limit for the contraction (minutes) and F is the strength being applied in the exertion, expressed as a proportion of the maximum strength of the muscle (or group of muscles) being contracted. The results obtained by Rohmert required him to impose two important restrictions on the formula: 1) the proportion of the maximum strength should be no lower than 15%, in which he coincided with Monod (1956); 2) the time limit could be no longer than 10 minutes. These two restrictions are evident in the shape of the curve obtained by plotting Rohmert's equation, shown in figure 2.7.

The influence of Rohmert's work on the subsequent research in this area has been quite strong and far-reaching, as mentioned already when reviewing the literature related with the cardiovascular responses to isometric exertion (section 3.1). Indeed, Drury and Spitz (1978) came to the same conclusion. They analysed a vast amount of information related to the topic of strength and duration of muscular effort, and identified no less than 12 studies in which Rohmert's equation played a central role.

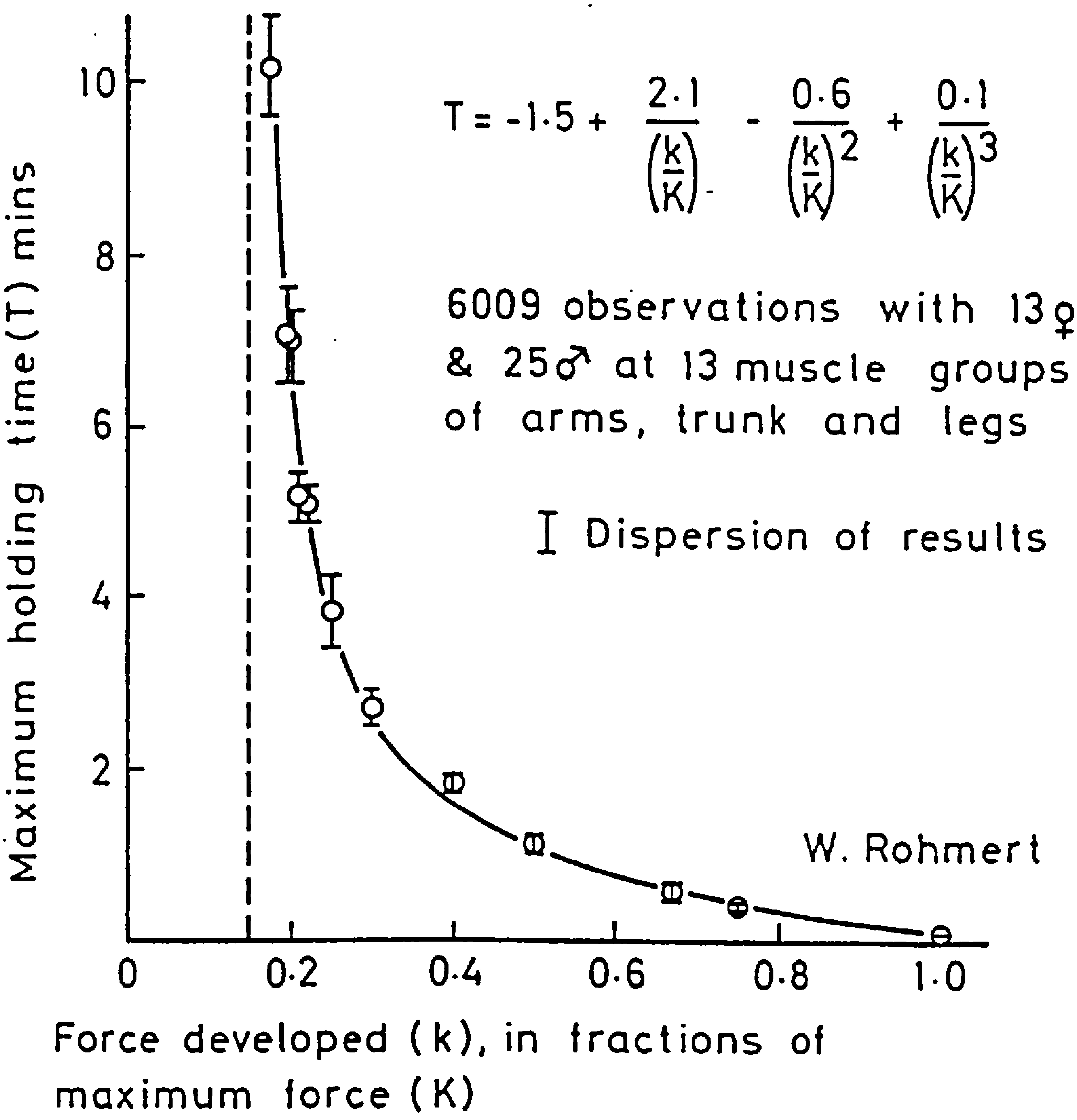


Figure 2.7 Relationship between the percentage of maximum strength and endurance in isometric contraction. (Modified from Rohmert, 1965)

2.10.3 An exponential model of strength and endurance

Deeb and Drury (1990) reviewed the methodology applied in the earlier studies that sought to establish the relationship between endurance time and strength of isometric exertion. They found that those studies agreed on two major accounts: 1) the existence of an endurance limit, that is a point in time beyond which the muscles cannot deliver the required force (either 100% MVC or a lower percentage from it); 2) the relationship between force and endurance time follows an exponential pattern. However, in the view of Deeb and Drury, the models generated by all those studies suffered from two major drawbacks: 1) the lack of a well defined criterion for identifying the endurance limit, and 2) the models are "based on weak assumptions and... (since) these models were typically fitted by eye to averaged data... the number of exponential components in each model varied depending on how many these researchers were able to fit to the curve". Consequently, Deeb and Drury developed a methodology to generate a model of the endurance to isometric exertion which, in their opinion, by taking advantage of the computational facilities nowadays existent, and that were not available at the time most of the early research took effect, should overcome those shortcomings.

The model describes the force produced by the subject over an extended period of time. That force is the larger of the one set as a target for the subject to deliver and the one the subject is capable of producing at any instant during the exertion. The latter force was defined by Deeb and Drury as the sum of exponential decays acting simultaneously upon the force-generating mechanisms. The model can be represented mathematically by an expression of the form

$$F = \max \left\{ F_{\text{required}}\text{, i.e. \%MVC required, } \sum_{i=1}^{P} a_i \exp[-k_i(t-t_e)] \right\}$$

where F (measured in kilograms) represents the force exerted by the subject; the constant a_i represents the strength (kilograms) generated by each of the possible mechanisms of force production available to the muscle; k_i is the fatigue rate (s^{-1}) that affects those mechanisms; t (seconds) is the total duration of the exertion and t_e is the endurance time (seconds). This expression is presented graphically in figure 2.8.

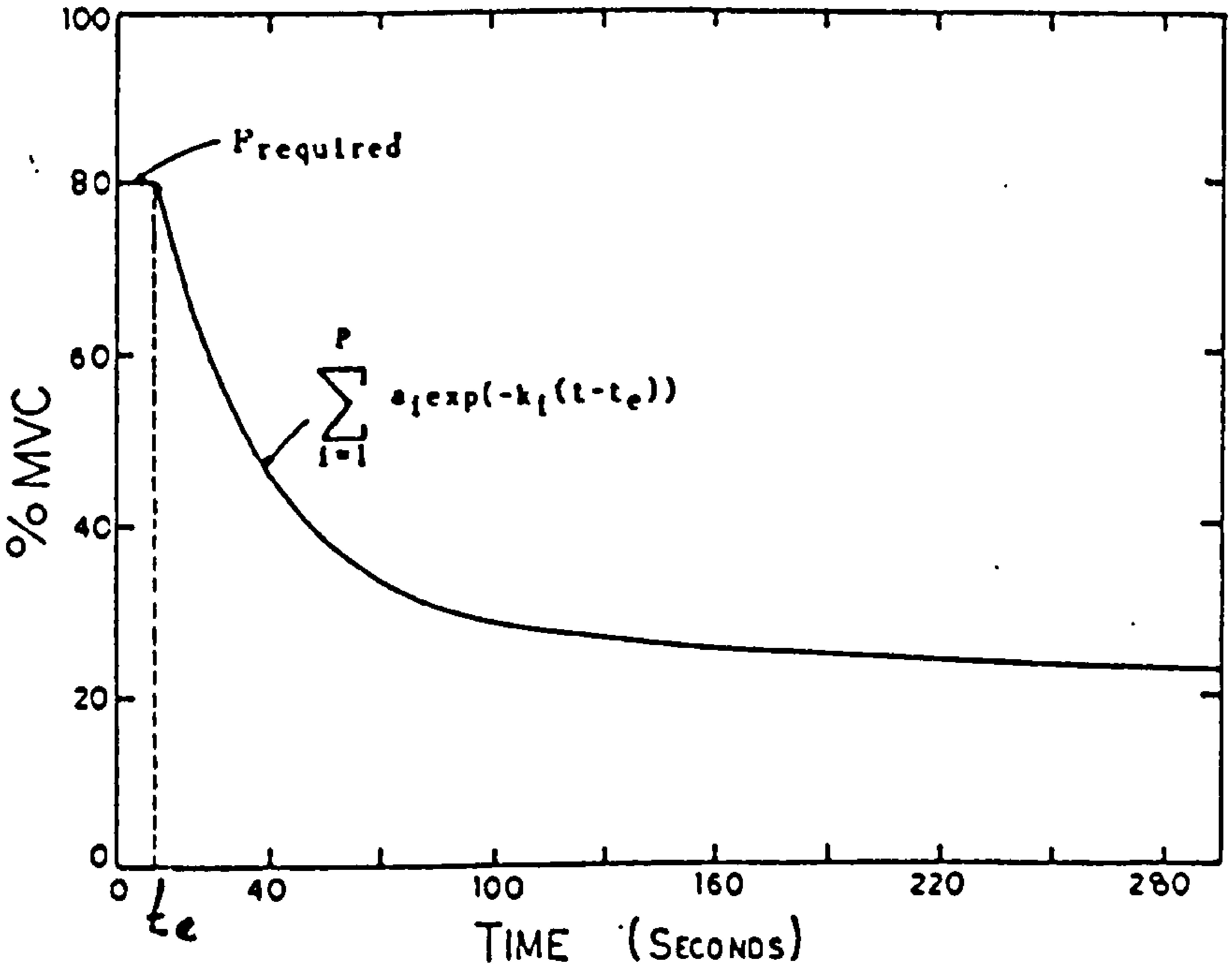


Figure 2.8 Graphical presentation of the generic model of fatigue in isometric exertion. (From Deeb and Drury, 1990).

To test their methodological approach, Deeb and Drury (1990) studied the behaviour of the force applied during the continuous contraction of either the biceps brachii or the quadriceps muscles. Two groups of ten male subjects each exerted forces equivalent to 20%, 40%, 80% and 100% MVC of the muscle. One of the groups was constituted by individuals aged between 20 and 29 years, the members of the other group were aged between 50 and 59 years. The duration of the contraction was fixed at five minutes. Drawing from the current knowledge about the fatigue process in isometric exertion, the authors proposed that the overall change in force during the exertion would be most likely represented by the combination of two exponential terms, that is P=2 in the mathematical expression of the model. Thus, the generic structure of the final model took the form

$$F_a = a_1 \exp(-k_1 t) + a_2 \exp(-k_2 t)$$

with a_1 and a_2 representing proportions of MVC that add up to the level of strength applied at the beginning of the exertion.

The method used to determine the value of t_e, the endurance limit, represented the major departure of this approach from earlier methodologies; it consisted in the iterative fitting, using a computer programme, of a non-linear function with the parameters a_1, a_2, k_1, k_2, to proposed values of t_e, drawn from the force and time data. The procedure was repeated as necessary until the total sum of squares for the function converged to a minimum, so determining the optimum values for the endurance limit and the four parameters.

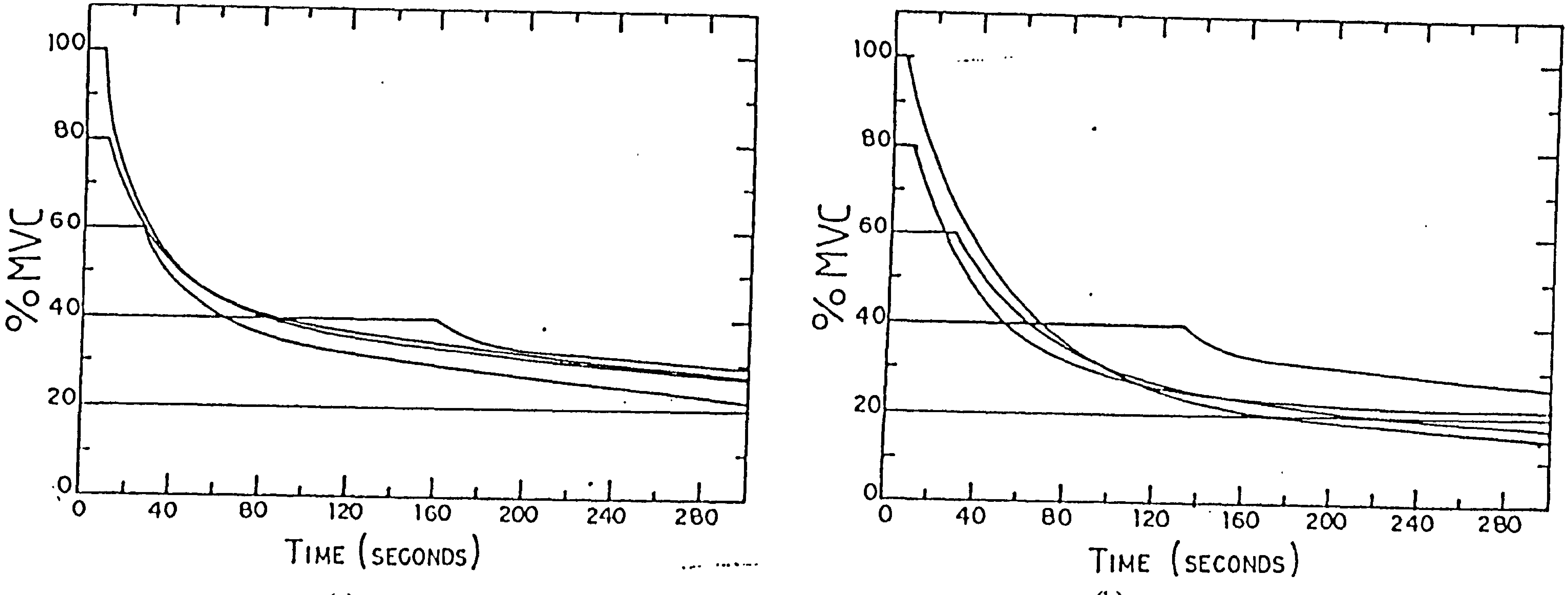
In a further development, Deeb et al (1992) suggested that the first exponential term in the generic model (see above) represents the portion of the exertion attributable to the recruitment of fast-twitch muscle fibres, and the second term corresponds to the usage of slow-twitch fibres. This suggestion followed from the knowledge that, in order to deliver high levels of force, the muscles recruit preferentially fast-twitch fibres and, as the force decreases by the effects of fatigue, fast-twitch fibres are gradually replaced by the more fatigue-resistant slow twitch fibres. Since biceps and quadriceps muscles are known to differ in their fibre composition, and the relative proportion of the two types of fibre in any given muscle may change with ageing, Deeb et al (1992) tested their assertions by analysing the data collected during the experiments carried out by Deeb and Drury (1990), applying the methodology developed by those authors and that has been described already.

Deeb et al (1992) used analysis of variance to assess the influence of the experimental factors age group, muscle group, level of exertion, and their interactions, on the values of the parameters a_1, a_2, k_1 and k_2. The results of those analyses showed that the age group did not affect significantly any of the parameters. The muscle group, the level of exertion, and the combinations of these two affected the parameters in different ways. They also found that the two-term model developed by Deeb and Drury (1990) accounted for 95% (on average) of the variation in the values of force observed during the exertions.

Regarding the involvement of the two types of muscle fibre in the generation of force, Deeb et al (1992) reported that both in the biceps and the quadriceps muscles, for the exertion at 20% MVC practically all the muscular force came from the recruitment of slow-twitch fibres, and the model explained the decrease of force with only the corresponding exponential term, $a_2\exp(-k_2t)$. When the exertion required strength levels of 40% MVC and above, the model incorporated both exponential terms, to account for the fact that fast-twitch fibres were recruited in proportions that grew linearly with the strength applied, whilst the proportion of slow-twitch fibres remained practically constant.

Deeb et al also found that the fatigue rates k_1 and k_2 reflected the difference in fatiguability of the corresponding muscle fibres, so that in the quadriceps muscle the strength provided by the fast-twitch fibres fell at a rate that was about 15 times that of the slow-twitch fibres, and in the biceps muscle that ratio was approximately 26 to 1.

Rather than use one single curve to depict the force changes that occurred during all the exertions they modelled, Deeb et al (1992) presented a set of curves for each muscle they studied, with one separate curve for each level of exertion, drawn using the values of the four parameters (a_1, a_2, k_1, k_2) averaged over the 20 subjects. Those curves are reproduced in figure 2.9. The observation by Deeb et al (1992) regarding the dominance of the slow-twitch element in the generation of force at 20% MVC is evident in the fact that at this level the force remained practically constant, making the value of a_2 equal to 0.2 (the percentage MVC being applied by the subject), and k_2 equal to zero.



(a)

(b)

Figure 2.9 Graphical representations of a mathematical model of fatigue in isometric exertion, which takes account of the muscular fibre composition. a) Plot of the change in force delivered by the biceps brachii muscle group; b) plot of the change in force delivered by the quadriceps muscle group. Reproduced from Deeb et al, 1992.

2.10.4 Rohmert's model for the calculation of rest allowances

Rohmert (1973) addressed the relationship between isometric exertion and rest. He found that the length of rest necessary in order to allow for a full recovery from the fatigue induced by an exertion period of length t/T_{max}, at a relative strength f/F_{max}, could be calculated with the expression:

$$\mathrm{R.A.} = 18 \bullet \left[\frac{t}{T_{max}}\right]^{1.4} \bullet \left[\frac{f}{F_{max}} - 0.15\right]^{0.5} \bullet 100$$

where R.A. stands for 'rest allowance' and is expressed as a percentage of t, the holding time which in turn is measured in minutes, and the strength of the isometric exertion is restricted to values above 15% of the maximum strength of the muscle. Using this equation Rohmert (1973) produced the series of smoothed curves shown in figure 2.10, which may be used to determine for a given combination of holding time (read as the ordinate, to a maximum of ten minutes) and relative strength applied (read as the abscissa), the length of the rest allowance (as percentage of the holding time) necessary to achieve full recovery, which is read from the nearest curve above the intersection of the corresponding ordinate and abscissa. The dashed vertical line traced from the value 0.15 on the horizontal axis is assigned a value of zero, reinforcing the assumption that the exertion of such strength or lower does not provoke fatigue and therefore does not require a rest allowance.

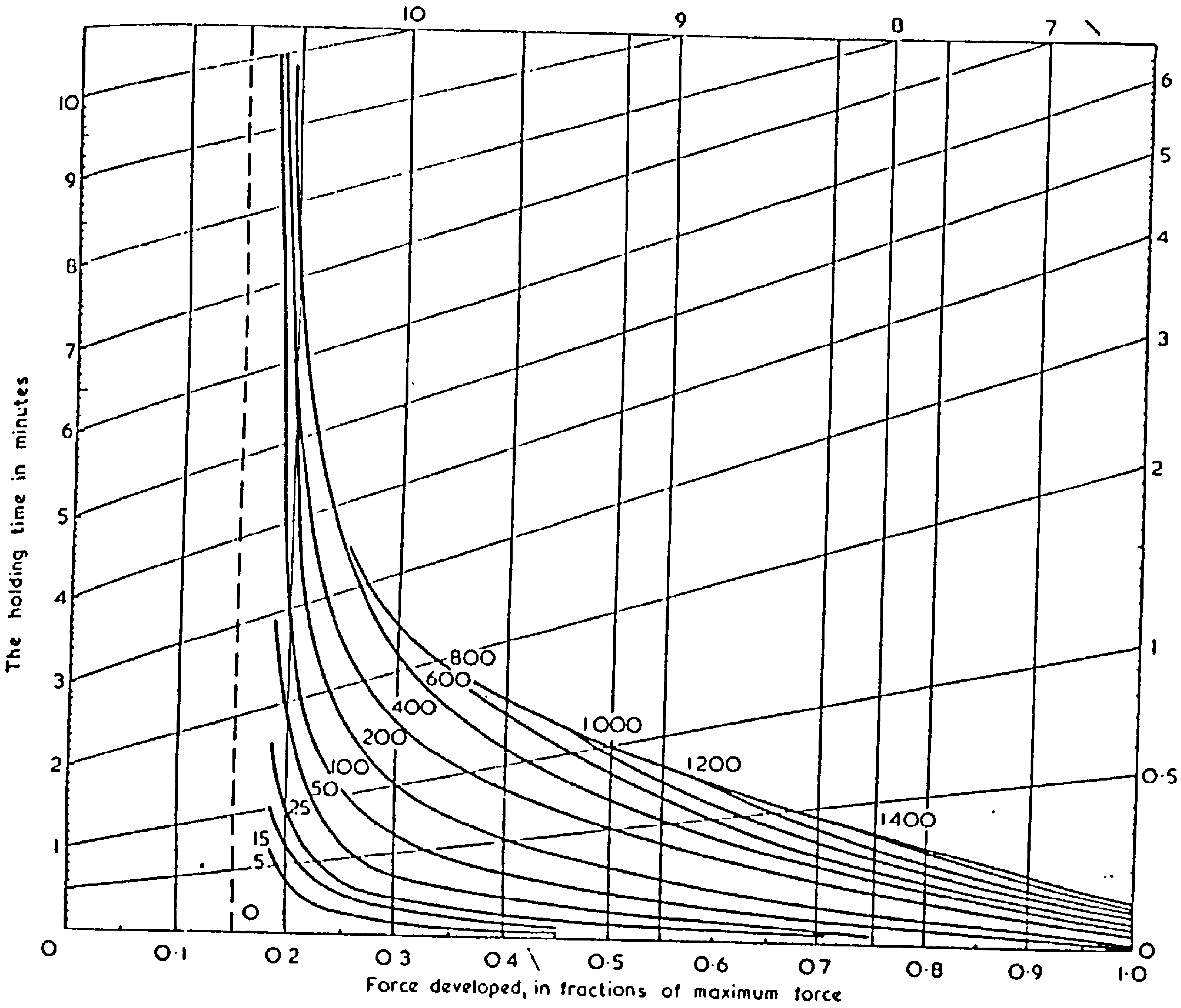


Figure 2.10 Percentage rest allowances for various combinations of holding forces and time. (From Rohmert, 1973).

The ultimate aim behind Rohmert's model for the calculation of rest allowances is to find the optimum combination of isometric work and rest, which will maximise the total amount of work that may be carried out without provoking cumulative fatigue. Monod and Scherrer (1964) recommended that, in order to perform the maximum amount of intermittent static work per unit time without provoking fatigue, the exertion period and the rest period should be of the same length, and the strength applied should be 40% of the maximal strength.

2.10.5 Evidence disputing the existence of a 'non-exhausting' level of force

The notion that the isometric exertion of a force lower than 15% MVC could not possibly lead to exhaustion and may be sustained for a long time, hours even, is central to the work of both Rohmert (1960) and Monod (1956). This, however, has been repeatedly questioned, particularly from the evidence gathered through the collection and analysis of electromyographic information. So, Davies and Pratt (1976) recorded the maximum endurance to handgrip exertions performed at 15% MVC, finding that their subjects could only sustain such level of exertion for between 3 and 16 minutes. Bjørksten and Jonsson (1977) studied the endurance to elbow flexion both for a continuous static contraction and for intermittent work and rest. They found that the highest force that could be sustained continuously for one hour was approximately 8% of the elbow flexor's MVC, although they could demonstrate signs of fatigue in response to exertion of only 5% MVC; on the other hand, work combined with rest could be sustained for one hour only if the average level of force did not exceed 14% MVC. From these results, they recommended that the force to be exerted continuously in an isometric contraction should be only between 2 and 5 percent MVC, and that long-lasting (a workday of 8 hours) intermittent or dynamic work should be limited to the exertion of between 10 and 14% MVC.

Jørgensen et al (1988) reported that approximately one hour was the longest endurance to continuous elbow flexion or knee extension at 10% MVC. Although exertions at 5% or 7% MVC could be sustained for one hour, this resulted in fatigue evidenced both in the subjective perception and by changes in the EMG; furthermore, one hour of exertion at 5% MVC resulted in a reduction of between 10 and 12% of the original MVC, contradicting Rohmert's (1973, p 92) assertion that "No reduction in maximum strength occurs, if the holding force is limited to 15% of maximum strength" . Jørgensen et al (1988) also reported on intermittent work, consisting of 1440 sequences of attempted pulling movements lasting for 10 seconds followed by 5 seconds' rest, performed over a period of 435 minutes; the average force required in the pulling movement was adjusted to be either 15% or 10% MVC. They found that 15% MVC provoked fatigue, which was evident very quickly (within the second hour of work), but work at an average 10% MVC did not provoke EMG changes indicative of fatigue.

Krogh-Lund and Jørgensen (1992) studied the changes in the EMG activity from biceps brachii and brachioradialis muscles that occurred when isometric flexion of the right elbow at 15% MVC was sustained to exhaustion. The subjects were eleven males in their twenties, who could sustain the exertion for an average of only 906 seconds, at the end of which there were clear signs of fatigue as detected by changes in the EMG signal. Krogh-Lund (1993) reported on the changes in the EMG signal from the elbow flexors that occurred as a consequence of isometric continuous exertion at only 10% MVC, sustained to exhaustion. The mean endurance to such level of force was 51 minutes, and again the characteristics of the EMG signal showed changes indicative of fatigue.

Caffier et al (1993) studied the EMG changes brought about by one hour of isometric contraction (or shorter, if the subject could not endure the whole hour) of the right biceps, with force equivalent to 4%, 8% and 15% MVC. They worked with 12 young male subjects; all 12 could reach the target time of one hour with the contractions at 4% and 8% MVC, but only 8 of them were capable of sustaining the contraction at 15 % MVC for the whole hour. However, even a load level as low as 4% MVC produced signs of fatigue that could be detected in the EMG signal, and this led the authors to conclude that quite possibly there is not a load level low enough as to allow an unlimited duration of contraction, as suggested by Rohmert (1960) and Monod (1956).

2.10.6 Differences in fatiguability of muscle groups

Rohmert (1960, 1973) also asserted that, as long as the holding forces are expressed in relation to the individual maximal strength, the relation between endurance and contraction force should be valid for all muscles, irrespective of individual differences. The findings from a series of studies also contradict this notion. Kroll (1968) studied a sample of 45 subjects, who were classified as being of low, middle or high strength. Each subject performed 30 trials of maximum isometric wrist flexion, and the results suggested that the weaker subjects tended to fatigue significantly more slowly than those of middle and high strength. Bjøkrsten and Jonsson (1977) studied the endurance to elbow flexion in four male and four female subjects, finding that female subjects tended to have a higher endurance limit than the males. By contrast, Takala et al (1993) reported that EMG responses to a test of maximal holding of the arm in extension with a 2 kg weight suspended from the wrist, suggested that male subjects tolerate larger changes than female subjects before reaching fatigue; this, however, did not reflect in terms of endurance time, which was on average the same for male and female subjects. Petrofsky and Phillips (1982) reported that the muscles used in handgrip were more fatiguable than neck extensor muscles. Sato et al (1984) found that shoulder abductors are more fatiguable than elbow flexors or knee extensors.

It is quite remarkable then that despite the ample evidence available at the time, contradicting the notion of equal fatiguability for all the muscular groups and for every person, in 1989 Kahn and Monod still wrote that the equation developed by Rohmert (1960) is applicable to all muscles and holds valid irrespective of the level of fitness exhibited by the subject. This is quite a contrast with the position of Rohmert himself. When he first put forward the model for the calculation of the maximum exertion time of a static force (already reviewed in section 2.10.2), Rohmert asserted that it was valid for all the muscle groups in the body. However, in a study where the model was used for the assessment of five working postures (Rohmert et al 1986), it was found that the model was appropriate mainly for simple postures that only involve the exertion of the muscles in the arms and shoulders, but for complex postures where muscles of the back, trunk and legs interact, the model cannot predict adequately the endurance to the postural loads so created. Rohmert et al concluded that in such conditions a new static *postural* (authors' italics) force model should be found for each particular posture, and it would be expressed by an exponential equation of the form $Y = A \cdot X^B$.

Thus, in coming to the end of this review of the results from the search for a unitary model to explain the relationship between the strength applied during an isometric exertion and the endurance to that effort, all seems to indicate that such goal is still some way out of reach. If anything may serve as evidence to back this assertion, the above cited conclusions reached by Rohmert et al (1986) and the results of the study by Deeb et al (1992) should suffice. These two sets of results point to a situation in which, although still counting with a basic structure on which to build a generic model, in the end there could be a different version of that model for each muscle group.

2.11 Models of fatigue and recovery in purely postural work

All the approaches to modelling the fatigue that results from isometric exertion reviewed up to this point have evolved from the study of contractions by a well defined muscle or group of muscles, which applied a level of force measured as a proportion of the maximum strength that those muscles could produce. However, such approach can hardly be extended to the study of fatigue that results from the holding of a whole posture, for two main reasons. The first one is that posture holding involves the activation of rather large groups of muscles, with possibly quite complex patterns of interactions that make it difficult to ascribe the development of fatigue to one of them in particular; the second difficulty lies precisely in finding an appropriate way to measure safely and accurately the maximum strength of all the muscles involved. Besides, if the interest centres around the postures adopted during the performance of a job, studied at the worksite itself, it is unrealistic to envisage the application of methodologies that are time-consuming and often require the use of equipment which cannot easily be taken away from a laboratory setting.

2.11.1 A model of recovery as a function of the duration of postural loading

Bearing in mind the considerations just expressed, Barbonis (1979) investigated the possibility of predicting the recovery from what he called 'postural work load', i.e. the muscular effort involved in the sole holding of a posture, based only on the knowledge of the length of time the posture has been held and the length of rest allowed following that exertion. He studied the course of the fatigue arising from the holding of five postures that imposed varying degrees of stooping. His methodology consisted of having the person perform a maximum holding, that is hold the posture until they reached their limit of endurance (defined as the moment when the person judged that discomfort or pain had become unbearable), then assigning a period of rest with a duration calculated as a multiple of the measured maximum holding time (MHT). Following this, he asked the person to hold the posture again to the limit of endurance. If this second holding lasted for as long as the first one, that should prove that the rest had been long enough to allow full recovery. Barbonis (1979) found that most of the recovery took place in the short term after the end of the initial exertion, although in some cases full recovery was not achieved even after a rest twelve times as long as the maximum holding time.

Figure 2.11 shows the plot of the overall relationship between rest and recovery as reported by Barbonis (1979). He also found that the relationship between the length of rest and the recovery followed the same pattern for the five postures he studied, despite the obvious differences in the extent of the load they placed on the muscles involved, and this he interpreted as evidence that "at least for the five postures examined... knowledge of the magnitude of the load on the various muscles contributing to the maintenance of the posture is not necessary for the assessment of postural work recovery times".

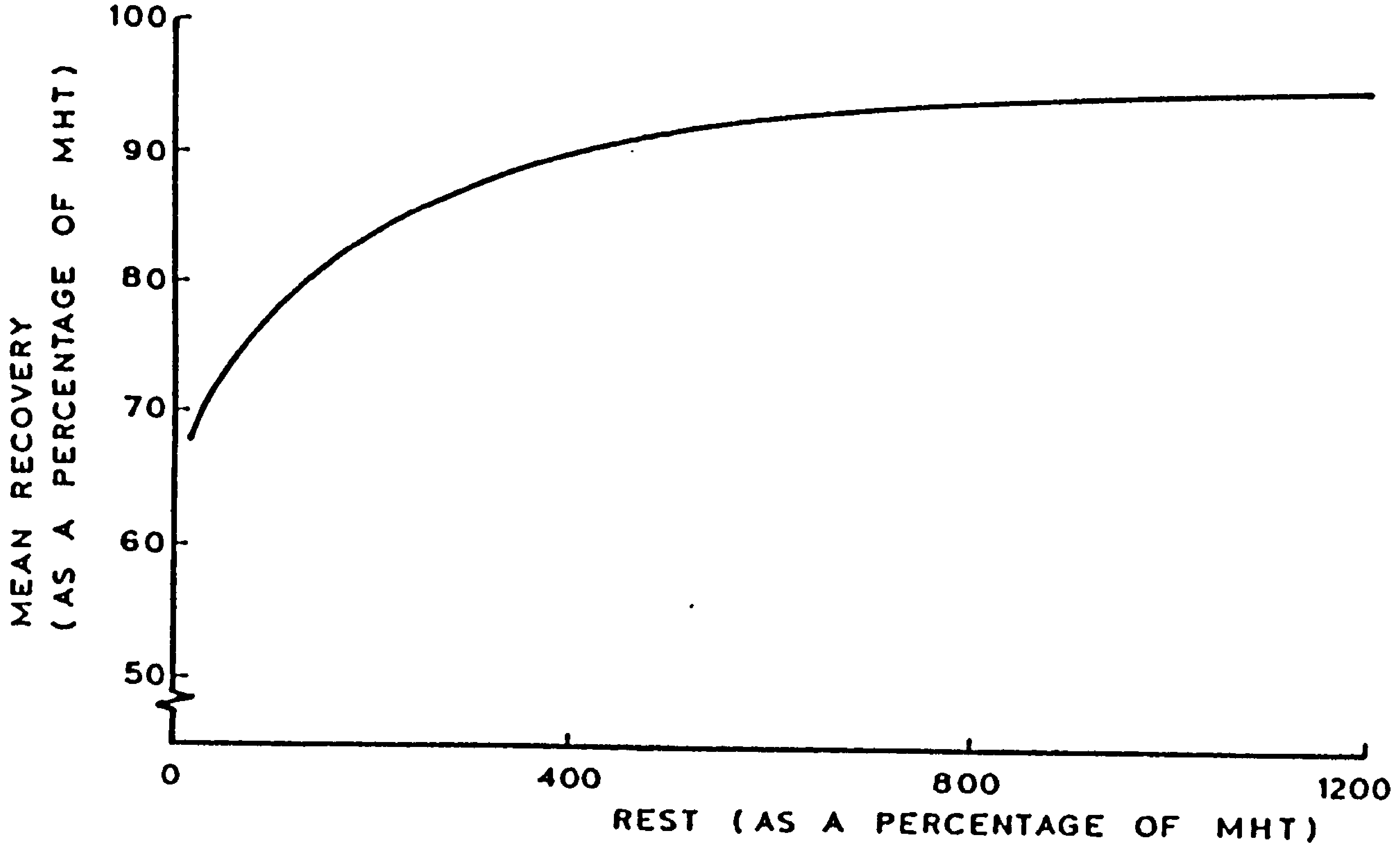


Figure 2.11 Degree of recovery from postural work after different lengths of rest, following a maximum holding time (MHT).

(Reproduced from Barbonis, 1979).

The possibility of predicting the recovery from pure postural exertion without having to know first the actual load borne by the muscles has an obvious appeal, particularly in those situations where it is likely that much of the physical demand on the person comes from that form of exertion, as may be the case in highly sedentary office jobs or work at an assembly line or where the arrangement of the work place forces the person to adopt awkward postures.

The work of Barbonis (1979), however, had at least three important limitations that needed to be addressed in order to enhance the applicability of his results to practical situations. First, there was the issue of the unfeasibility of the work and rest regime applied in the development of the model, since it is difficult to envisage a practical situation where people are asked to hold a posture to their limit of endurance, and then allowed to rest for at least the same length of time. Second, the model was developed from the analysis of data obtained from single combinations of work and rest, and this raises the question of whether the events he observed during those single combinations would occur in the same way were the experiment to be carried out several times, on separate occasions. Third, although an attempt was made to assess the development of fatigue through the changes in heart rate, this showed a poor correlation with postural load. Therefore, the question persisted: is pure postural exertion a fatiguing activity on its own right, and if so, what physiological variable is the one that best reflects this?

2.11.2 Milner's model of recovery as a function of the maximum endurance to postural loading

In a subsequent laboratory study, Milner (1985) addressed the three issues mentioned above, and at the same time expanded on the main findings reported by Barbonis (1979). The basic aim of Milner's work was to develop a mathematical model to predict the recovery to be achieved by a person after undergoing a series of combinations of postural work and rest, their lengths shorter than or equal to (but never longer than) that of the maximum holding time for the posture being held. That posture was one of the five on which Barbonis (1979) had tested earlier. It required the subjects to place their hands at a height equal to half their shoulder height, and at a distance to the front equal to their arms reach.

In an attempt to establish what physiological variable could best reflect the effects of the postural loads, Milner assessed the cardiorespiratory responses to the postural exertion. This he did by measuring the changes in heart rate, blood pressure, composition of air expired and blood levels of lactate, and relating those changes to the duration of the posture holding exercise. Milner carried out a short study to compare the responses of six subjects to two separate maximum holding times and to a third exertion that lasted for half of the first maximum holding. The results showed that, in contrast with the findings of Barbonis (1979), heart rate exhibited a trend to increase linearly with the passage of the holding time; blood pressure also increased with the holding time, although the relationship was significant only for the systolic phase. Finally, neither the composition of the air expired nor the blood levels of lactate had a significant relationship with the length of the holding exercise.

In his main study, Milner (1985) started by measuring the endurance of the subjects to the postural exertion, by asking them to hold the posture for as long as possible, until the discomfort became unbearable. The rating and subsequent quantification of discomfort were done following the procedures developed by Corlett and Bishop (1976). Next, the subjects were required to perform a series of experiments in which they first held the posture for a length of time that was a portion of the maximum holding time, then they rested for a time that could be either equal to or half the length of the exertion just performed, and finally they held the posture again, on this occasion until reaching the endurance limit. The length of this second holding represented the portion of the maximum endurance left after the performance of the combination work/rest; Milner called it recovery and noted that it could be expressed either in units of actual time or as a percentage of the maximum holding time.

The rationale behind the use of the term 'recovery' when referring to the length of time that a subject can hold a posture after an earlier holding followed by rest, is that such length of time measures the capacity for the exertion that has been restored by the rest pause. In his experiments, Milner used initial holdings of duration equal to 25%, 50%, 75% of the original MHT, and a second MHT; these were combined with rest periods that lasted for half of or an equal length to the holding just performed.

The next stage in Milner's work was to search for a mathematical expression of the relationship between the values of recovery (REC) attained following the first holding time (HT) and the rest (R), all of which were related to the length of MHT. He ended up with the following equation:

$$\mathrm{REC} = (\mathrm{MHT} - \mathrm{HT}) + \mathrm{HT} \bullet e^{(-0.164 \bullet \mathrm{HT}/\mathrm{R})}$$

which will be referred to as 'Milner's model' for the remainder of this thesis. The two-term structure of the equation reflects the fact that recovery happened as a two-stages process, the first of them a quick, steep change that occurred in the early part of the rest period, accounted for by the first term in the equation. The second term reflected both a slowing of the recovery rate and the fact that for any given length of initial holding, a shorter rest achieved a lower recovery. Figure 2.12 illustrates the curves fitted by Milner to the results obtained during his experiments, using the equation given above. The points placed on each curve represent the level of recovery averaged over the measurements for 24 subjects.

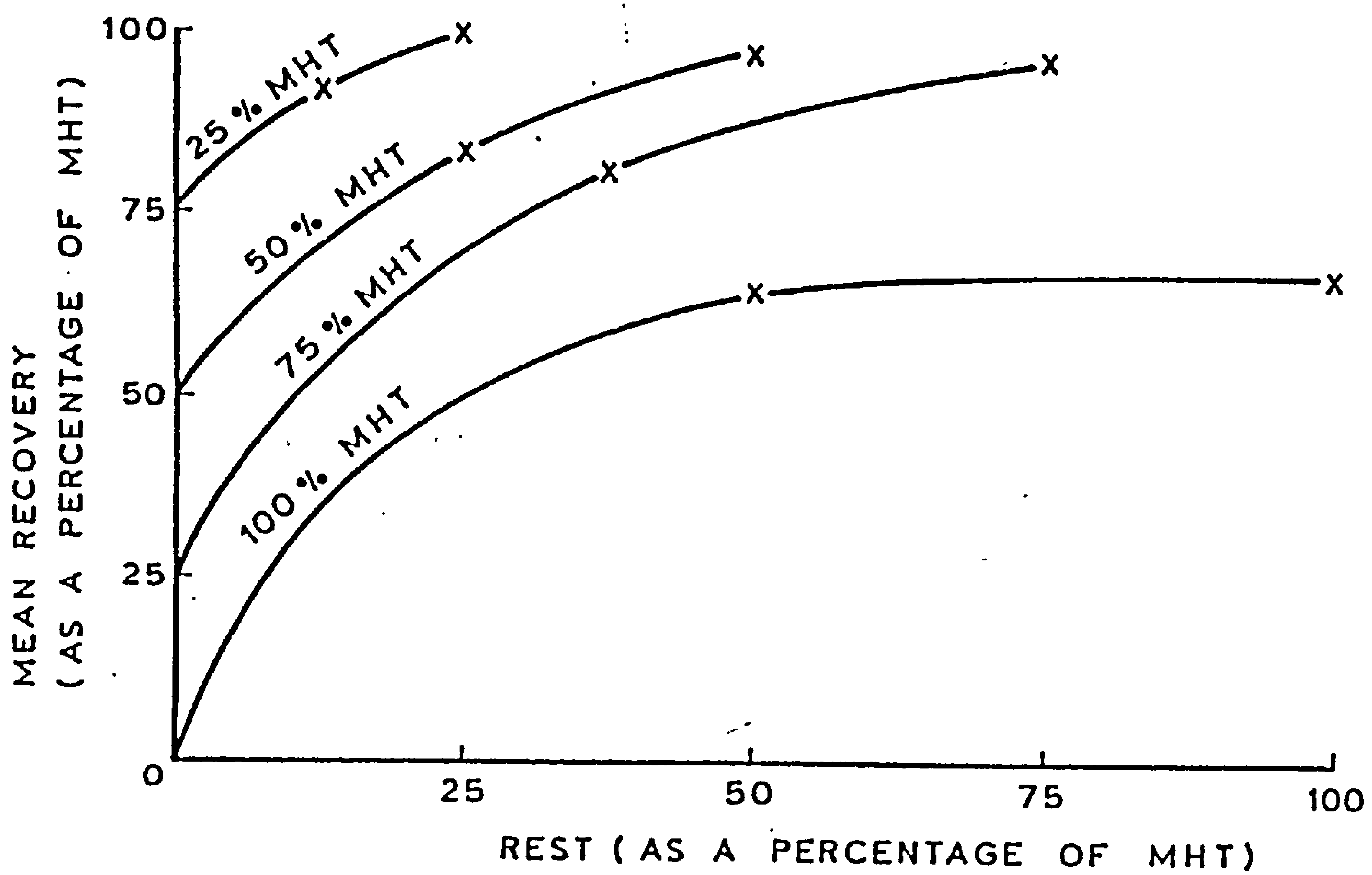


Figure 2.12 Graphical representation of the relationship between recovery, duration of holding time and rest, as expressed by Milner's model. (Reproduced from Milner, 1985).

2.11.2.1 Attempts at widening the application of Milner's model

Milner attempted to use the equation model to predict the level of recovery that might be achieved following more than one single combination of work and rest. To this end he asked six subjects to complete as many sequences holding/rest/holding as they could, until they were unable of reaching a recovery level previously set as a target (naturally, unknown to the subject). Milner found that, in its original formulation, the equation tended to underestimate the levels of recovery (i.e. measured > predicted), with the extent of the difference growing as more combinations were completed by the subject. Milner attributed the presence of the shortfall to the influence of the rest periods included in the earlier combinations. To compensate for this effect, he modified the model by including an additional exponential term. The transformed equation took the form:

$$REC_{i+1} = REC_i \bullet e^{-a\left(\frac{R}{MHT_0}\right)} + MHT_0 \bullet \left[1 - e^{-a\left(\frac{R}{MHT_0}\right)}\right] - HT \bullet \left[1 - e^{\left(\frac{-0.164 \bullet HT}{R}\right)}\right]$$

where MHT_0 is the length of the original maximum holding time.

To find the value of the constant 'a' incorporated in the exponent of the first two terms, Milner applied a trial and error approach. He compared the observed levels of recovery with those predicted by the equation using different values of 'a', looking for a value that minimised the difference (measured - predicted). However, there was not a value that fulfilled that criterion in every possible circumstance. With 'a' equal to -0.5, the equation predicted with acceptable accuracy only for the sequences that combined two holdings to exhaustion separated by a rest of duration equal to the first holding. For the combinations of shorter holding times, the differences were still significantly large. This led Milner to conclude that far more research was needed before it could be confidently said that the modified equation was in fact the best option.

Milner also tested the applicability of the original equation to a posture different from that used to develop it. This was an upright posture with the arms raised to the front, placing the hands at the subject's height and at a distance equal to arm's reach. Milner found that, in those conditions, the recovery predicted by the equation was consistently larger than the measured recovery, although the differences appeared to be non-significant. Nevertheless, the model has still to be tested for its applicability to a wider variety of postures, particularly those which are a more frequent occurrence in the workplaces where the circumstances impose postural constraints on the worker.

2.11.3 A work-rest model for purely postural exertion

Dul et al (1990, 1991) have proposed a model which may be used to determine, in case of purely postural static work, the combination of work and rest periods that will result in the lowest overall level of fatigue. The authors stated that the model may be applied to practically any combination of work and rest lengths, and to any number of cycles. Their central thesis was that the endurance to a posture is determined by the so-called critical muscle group, the one that is placed under the heaviest stress by the posture in question, which may be identified by measurements with EMG, biomechanical analysis or even psychophysical methods. However, Dul et al (1990, 1991) did not provide an equation (or equations) for the model, but affirmed that it was based on equations previously developed by other authors. Thus, the relationship between the force exerted by the critical muscle group and the longest time that force may be sustained was taken from Sjøgaard (1986), who in turn had extrapolated to a duration of 8 hours the relationships established originally by Rohmert (1960), Bjøkrsten and Jonsson (1977) and Hagberg (1981b). The endurance calculated from that relationship yields the value of the absolute maximum working time, or t^0_{max}. The model developed by Milner (1985) was used by Dul et al (1990,1991) to calculate the proportion of the maximal endurance to the posture which should remain after the completion of each combination of work and rest (recovery in Milner's model), and to this they called maximum working time, or t_{max}. The authors assumed that, as suggested by Rohmert (1960), both the model of endurance and the model of fatigue and recovery were valid for all the critical muscle groups, irrespective of their location on the body.

The final outcome of the model is called 'muscle fitness' (f), which is the conceptual opposite of muscle fatigue, and is expressed as the percentage ratio of the maximum working time (t_{max}, calculated using Milner's model) to the absolute maximum working time (t^0_{max}, obtained from Sjøgaard, 1986) that is $f = (t_{max}/t^0_{max}) \bullet 100\%$. However, rather than calculated as this ratio, muscle fitness is estimated from the discomfort ratings returned by the subjects on a

10-point scale (Borg, 1982). This is based on the assumption that, for groups of subjects, muscle fitness and discomfort ratings are linearly related (with slope = 1.0) so that, for example, when a subject rates the discomfort as 5 (mid-way on the scale) the muscle fitness is 50%.

The difference between predicted and observed values of muscle fitness varied depending on the magnitude of the muscular effort involved in the postural exertion, it was reported as within 10% in the 1990 paper, but in the 1991 paper they mentioned that it could be as large as 30%. Figure 2.13 shows the results obtained by Dul et al (1991) when they applied the work-rest model to the holding by ten subjects of a posture that loaded the shoulder region. Part a) depicts the results from 5 cycles work/rest of duration 2 and 4 minutes, respectively; the corresponding lengths of the 6 cycles work/rest represented in part b) were 1.5 and 3 minutes.

Dul et al (1990, 1991) also have developed a computer programme for the application of their model, which offers the possibility of calculating either the level of muscle fitness that will be left following a number of work/rest combinations of known length, or the length of rest that needs to be combined with a given work time, in order to keep the level of muscle fitness above a certain value.

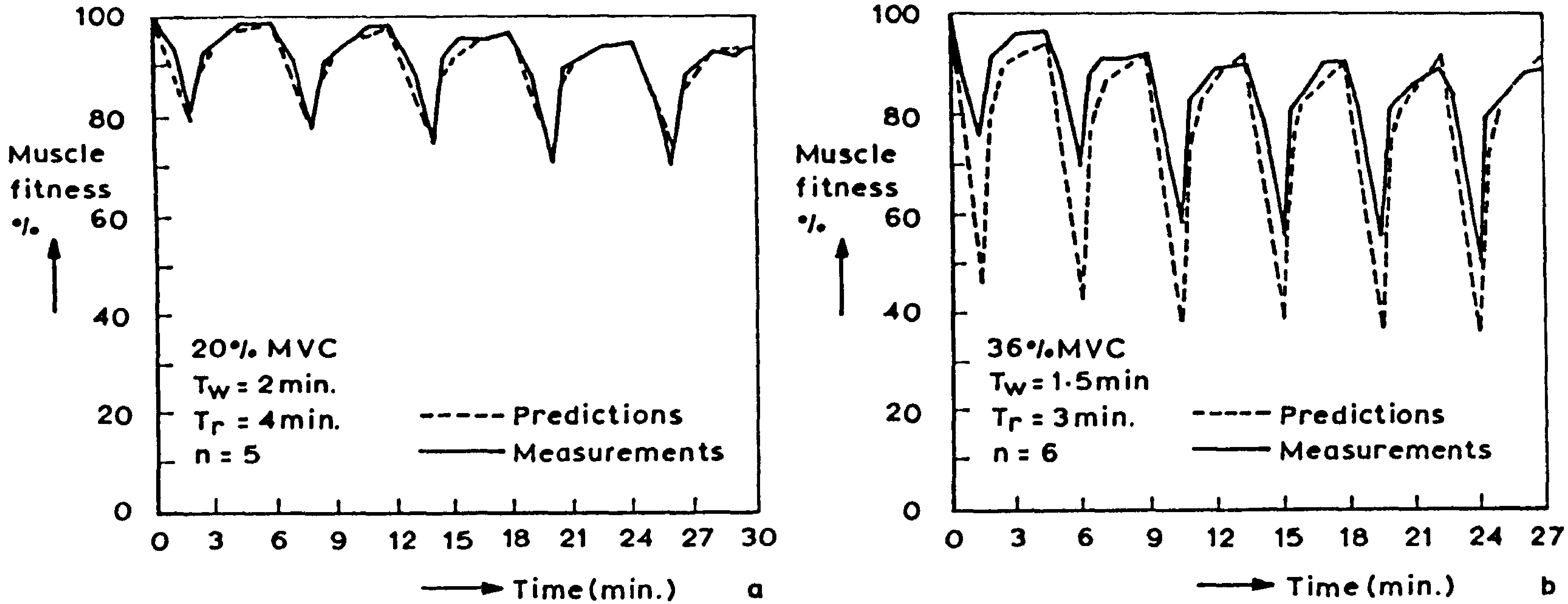


Figure 2.13 Results obtained from the application of the work-rest model for purely postural exertion. Example of the best (a) and worst (b) agreement between model prediction and recorded muscle fitness. (Modified from Dul et al, 1991).

2.11.4 Application of the work-rest model for the prevention of excessive postural discomfort

Based also on the notion that discomfort ratings on the 10-point scale (Borg, 1982) and the endurance to purely postural loads are linearly related, Dul et al (1993) have extended their work towards the proposal of guidelines about the maximum permissible duration of a static postural effort. The central thesis in their proposal is that by preventing the appearance of excessive discomfort, this will eventually reduce the likelihood of the development of musculoskeletal trouble. After analysing the results from a series of studies about the relationship between endurance to postural loads and discomfort ratings, Dul et al (1993) have proposed that, in order to avoid excessive discomfort, the exposure to postural loading should be limited to the time it takes for the discomfort ratings averaged over a group of workers to reach the value 2 ('weak' discomfort) , since this would ensure that at least 95% of the individuals will not reach the level of 'strong' discomfort, which corresponds to a rating of 5.

Mathiassen and Winkel (1992) pointed to what they saw as serious pitfalls in the modelling approach taken by Dul et al (1991). In particular, they referred to the unwarranted extension of Milner's model to circumstances quite different from those where it was originated, and to the important effect of interindividual differences in endurance, which Dul et al (1991) apparently disregarded altogether. Mathiassen and Winkel (1992) also stressed that when it comes to the muscles of the shoulder, very little is known about their endurance to isometric exertion. Quoting results from their own research, Mathiassen and Winkel (1992) highlighted the fact that the endurance limit appears to bear little relation to the physiological changes that occur during isometric exertion, particularly when it is performed intermittently, nor does it seem to be related to the phenomena occurring during recovery. From these observations, Mathiassen and Winkel (1992) concluded that the endurance limit is not a convenient way to measure the loads created by postural exertion, and by extension, the risk of musculoskeletal disorders; they stated that physiological measurements should be preferred in all circumstances.

Dul et al (1994) have addressed some of the criticisms expressed by Mathiassen and Winkel (1992). They stressed above all that the latter authors appeared to have misinterpreted the intention of the model presented in Dul et al (1991), for whilst they saw it as a means to determine, at a group level, the extent of the remaining endurance capacity (REC) following either constant or intermittent postural exertion, Mathiassen and Winkel (1992) analysed its use as a predictor of endurance limits, which are actually used only as an input for the model. Citing again the linear relationship between discomfort ratings and the endurance to postural loads (which although not true for every individual, holds at group level), Dul et al (1994) justified the use of discomfort ratings as an independent measure of the stress created by postural demands, which may be (or maybe not) reflected in short-term physiological changes. They also stressed that some of the limitations of the model pointed out by Mathiassen and Winkel (1992) stem from the limited amount of information on the topic currently available, particularly in reference to the long-term responses to isometric loading. Mathiassen and Winkel asserted that, because of its poor correlation to changes in physiological variables widely accepted as indicators of muscular function (concentration of lactate, EMG activity, maximal strength of the muscles), the endurance limit is not a valid indicator of the physiological state of the individual submitted to postural exertion. To this, Dul et al (1994) countered by noting that the correlation between the physiological parameters used by those authors were not particularly better, and their usefulness could also be called into question, since none of them appeared as the best possible indicator on its own.

Thus, the arguments put forward by Mathiassen and Winkel (1992) and by Dul et al (1994) offer perhaps the best possible illustration of the fact that there are two aspects to the issue of muscular fatigue still awaiting to be satisfactorily solved. One of them is whether fatigue is best assessed by looking into the physiological changes or by resorting to the subjective perception, the other one is whether it is possible to find an adequate expression for the relationship existent between them.

2.12 Conclusions

This review exercise on the fundamental issues connected with the fatigue provoked by isometric exertion has shown, first and foremost, that there is still much to be learnt about the phenomena that occur during the performance of muscular exertion of low intensity and relatively long duration. In fact, the bulk of the experimental evidence in the field of static exertion comes from studies carried out in exactly the opposite circumstances: isometric contractions where the muscles are activated either to their maximum capacity or to an important proportion of it, which determines that the exertion may be sustained only for very brief periods. However, the information here reviewed has made evident that, despite extensive research, even the most fundamental issues are still a matter of controversy. This means that, to date, not a definite answer has been found to three elementary questions: 1) Is it in the brain or in the muscles and ancillary structures that fatigue is actually sensed? 2) Is there a single means to measure the extent of fatigue? 3) How do we recover from fatigue?

Clearly, although some of the procedures in it employed have been refined to an amazing degree, the measurement of fatigue through changes in the physiological status of the individual, particularly that of the cardiovascular system, is still a matter of controversy. It is frequent to find studies that, despite having been conducted in what appeared to be similar conditions and to closely related purposes, their authors report widely divergent results and call on quite different mechanisms to explain the changes observed.

The analysis of the changes that occur in the electrical activity generated by the muscular activity is another form of measurement that attends to the physiological conditions of the human being whilst performing isometric exertion. However, as it was stressed in the corresponding section of this review, it is a procedure that if not applied properly will yield highly misleading results.

Even though fatigue is a phenomenon linked to physiological changes, it is also true that a complete understanding of the phenomenon will only be achieved by addressing the subjective perception of those changes. However, there has been a very strong tendency among researchers to consider separately the physiological (some authors call it objective) from the subjective aspects of fatigue; there is very little information regarding how those two dimensions are related. In consequence, many issues remain to be thoroughly searched and settled before it may be said with absolute certitude that fatigue arising from isometric exertion may be precisely defined and measured attending either to the changes in the physiological state of the subject, or to their perception of how fatigued they are or, preferably, to a combination of both criteria.

Having devoted a large proportion of this review to the matters concerned with the role that the loads created by the prolonged holding of postures, and in particular arm abduction, play in the development of disorders that affect the musculoskeletal system, two conclusions seem pertinent. On the one hand, it may be concluded that nowadays there is enough evidence to affirm that posture is in fact a major factor in the appearance of M-S disorders. However, despite all that evidence and the vast amount of advice dispensed by ergonomists about the convenience of taking more care of posture-related issues, the problem is still well evident and it would be unrealistic to say that it is going to be easily solved.

On the other hand, especially over the last 20 years, the isometric exertion of the muscles in the shoulder area has drawn much attention, with the number of laboratory and field studies growing steadily, most of them remarking on the sheer complexity of the relationships between muscles and other structures in the shoulder. However, the few among those studies that have addressed the issue of the fatigue provoked solely by the holding of the arms in abduction, have looked into postures that bear little resemblance to those occurring in the occupational settings. It is therefore necessary to widen the knowledge in this particular area.

Regarding the research efforts in looking for a model of the relationship between the strength applied during an isometric exertion and the endurance to that effort, the outlook is that, rather then a single model, there will be a series of basic models which, with adequate adjustments, will apply to different groups of muscles.

The last section of the review, in particular, has shown that when the research moves into the study of whole postures, an area where the researcher cannot exert a strict control on the main variables, and the attempt is made to evaluate the effects from fatigue based on subjective perception in addition to (or instead of) the measurement of changes in physiological phenomena, then it becomes quite difficult to assess the relevance of the results from investigations on forceful exertion limited to a single muscle or group of muscles. However, in the progression from rather narrow laboratory-based research efforts onto the wider scope of problems that arise in the actual workplace, the study of whole postures and their effects on the musculoskeletal system is one of the main issues that need to be addressed.

Only one study (Milner, 1985) has addressed the possibility of modelling the course of fatigue and recovery from postural exertion without having to establish first the force being applied by the muscles most heavily involved in the posture studied. However, when the model was tested for its applicability to postures other than the one used to develop it, the results were inconclusive, as were those from the attempts to extend the modelling from a single combination of work and rest to multiple cycles. Therefore, it is important to submit the model to further testing, using postures that in addition to differing importantly from the one studied by Milner, also represent frequent occurrences in everyday working life.

The line of enquiry started by Barbonis (1978) and continued by Milner (1985), looking into the development of muscle fatigue without the need to know the loads acting on them, has an obvious appeal from the practical point of view, since it would mean dispensing with complex and time consuming measurement techniques. However, the heavy reliance of this approach on subjective judgement to determine the growth of discomfort and the reaching of the end point of the exertion, leaves it open to criticisms like those expressed by Mathiassen and Winkel (1992). There is an obvious need to reinforce the credibility of this methodology by relating the subjective perception to changes in physiological variables. Considering the difficulties encountered by Milner when he tried to use the response of heart rate as the means to assess the physiological load created by posture holding, and the inconsistent results this often produced, it is important to look into other forms of assessment which could reflect more specifically the course of fatigue as it happens in the muscle itself, the obvious candidate for that role being the measurement of electromyographic activity.

Finally, testing and extending the model of postural fatigue and recovery proposed by Milner (1985) acquires even more relevance because of its incorporation into other models, which are now being considered as the basis for far-reaching proposals in respect of the postures as a work-related stressor and the probable cause of musculoskeletal disorders (Dul et al, 1991, 1993, 1994).

ERRATUM: Page 114 is missing, due to a discontinuity in page numeration

CHAPTER 3

OBJECTIVES AND METHODOLOGY

3.1 Wide-scope objectives

The review of literature presented in the last chapter has shown that the subject of muscular exertion due to purely postural demands still remains relatively unexplored, and there are in it some basic issues where the current knowledge presents important gaps which need investigating. An improved understanding of this matter should eventually help to reduce the likelihood of people suffering injuries to their muscular and skeletal systems as a result of the postures they adopt during the performance of their occupations. This investigation aims at making a contribution to that far-reaching objective.

Concretely, the work presented in this thesis has dealt with the maximum endurance to postures that create loads on the muscles of the shoulder by holding the arms abducted and unsupported. It has explored in detail the possibility of extending to those postures a modelling approach which intends to explain the relationships between the onset of muscular fatigue whilst holding a posture, the length of time allowed for rest, and the performance of subsequent exertions.

3.2 Specific objectives

The concrete issues to be addressed by the present investigation have been mentioned already in chapter 1. In keeping with the statements expressed in pages number 7 and number 9, the following specific objectives were set:

1) To test the assertion made by Milner (1985) that a model to predict levels of remaining endurance to static postural work, developed from observations on a single standing and bent-forwards posture, is still valid when applied to other postures. In addition, since the model was derived from data obtained in a study of male subjects only, this research tested whether the model would also apply to female subjects;

2) to test how repeatable is the maximum holding time for a posture, which is assumed a valid indicator of the endurance to the loads created by that posture;

3) to evaluate the effects that variations in the posture and the gender of the subjects have on the maximum holding time;

4) to establish how the subjective perceptions of fatigue develop during the course of maximum holding times, and the way they are affected by postural variations and gender of the subject;

5) to assess the presence of muscular fatigue as indicated by changes in the electromyographic signals, to investigate the influence of the experimental conditions on their nature and extent, and to look for the possible relationships between those changes and the subjective perception of fatigue;

6) to assess the length of time over which the electromyographic signs of fatigue will persist following postural exertion of maximum duration.

3.3 Outline of the experimental work

The experimental work proceeded in two main stages. The first stage was devoted to fulfilling the first of the specific objectives, that is testing the suitability of Milner's model to conditions different from those where it originated. This was done by comparing the extent of recovery predicted by the model under test with recovery observed in a sample of subjects who held an upright standing posture with both arms abducted. Because of limited time availability, the sample used in this experiment was smaller than that used by Milner (1985) in his study, but special care was taken to recreate the experimental design he applied, so that the differences between the two studies were only those introduced on purpose. Details of this experiment, and the corresponding results, are given in chapter 4.

The second stage of the work constituted the main experiment. In this, a sample of five male and five female subjects, none of whom had taken part in the first stage, provided the information necessary to accomplish the remaining five specific objectives. The core work in this second experiment consisted of measuring, on three separate occasions, the maximum endurance to the muscular demands created by three upright postures with the arms abducted at three angles, including the one used in the earlier experiment. The main details of the experimental procedure followed to carry out the measurement of the maximum holding are described in this chapter. The results of the measurements, and the statistical treatment of those results to assess the repeatability of the endurance to postural loading (which constitutes the second specific objective) and the effects on it of the abduction angle and the gender of the subject (third specific objective) are presented in chapter 5.

At regular intervals during the holding of the posture, the subjects rated the strength of the discomfort or fatigue they were experiencing and this information then used to fulfil the fourth specific objective, that is to study the effects of the abduction angle and the gender of the subject on the perception of fatigue. Chapter 6 contains the details of these experiments and the main results.

Simultaneous to the monitoring of fatigue development through the subjective perception, electromyographic signals from 3 superficial muscles in each shoulder were collected. In order to fulfil the fifth specific objective, those signals were subsequently analysed, looking for the presence of changes indicative of the appearance of muscular fatigue during the course of the posture holding. The way of accomplishing the sixth specific objective of this research (assessing the persistence of signs of fatigue following the posture holding) was by comparing the characteristics of electromyographic signals obtained in similar conditions before and after the postural exertion. To obtain those signals, the subjects were asked to perform a series of movements designed to activate each one of the muscles from which EMG signals were collected. The subjects carried out those movements immediately before they held a posture to the limit of their endurance and five minutes after they had reached that limit. The procedures followed in the collection and analysis of the EMG signals and the corresponding results are presented in chapter 7.

3.4 Methodology

Some of the procedures were common to both main experimental stages that constituted this investigation. Therefore, rather than duplicating the description of those procedures, the methods used will be described in this chapter, although some specific details will be kept for the later chapters, when they will be presented in a more appropriate context.

3.4.1 Recruitment of subjects

Since the investigation being reported in this thesis consisted of trials carried out by human subjects, it is fitting to open this presentation of the methodology applied during the study by describing the procedures followed to select those subjects and to provide them with information about the experiments in which they took part. However, the relevant characteristics of the subjects will not be presented at this point, these will be kept for chapters 4 and 5, where the experimental work performed in each of the two stages of the study will be described in detail.

All the subjects were full-time students at Nottingham University. They volunteered their participation in the study by responding to posters that were placed in several places around the main campus. The main criteria applied in their selection were that the subjects should be aged between 18 and 24 years and should be in good general health, in particular, free from any kind of musculoskeletal trouble or complaint. In addition, they should not be engaged, either by work or leisure, in any sort of activity that could place their upper arms under a heavy strain. The subjects were paid a small fee for their participation in the study.

Every person who expressed interest in taking part in the experiments was given a screening questionnaire aimed at finding out about their current health status, whether they held a job or pursued leisure/sporting activities by which their upper limbs could be subjected to heavy strain. A small number of volunteers were rejected at this stage. Before being actually recruited, the potential subjects were given a full and detailed explanation of the aims of the investigation and the procedures they would be subjected to. If they decided to participate, they were handed an information package to take with them. The package contained the same information the subjects had just listened to, but the subjects were invited to read it, in order to help reinforcing their understanding of the investigation. The subjects were also told that, if they felt it was necessary, they should ask for any further clarification. Copies of the screening questionnaire and the information package (one for each of the two experimental stages) are included in appendix A.

After they had accepted to take part in the study, and before they were submitted to any experimental procedure, the subjects were asked to fill in a questionnaire stating their current state of health and to sign a form of consent. Copies of the health questionnaire and the consent form are also included in appendix A.

3.4.2 Experimental postures

Arm abduction is an action so important in everyday life that it is nearly impossible to think of any major arm movement in which it does not play a significant part. However, the skeletal and muscular structures of the shoulder may come to harm if the arm is repeatedly abducted to an extent beyond the natural, non-stressful range - 90° in active mode, 120° in passive mode, according to Lucas (1973)- or if a lower degree of abduction is held for a long time. It has been previously mentioned (chapter 1, section 1.6) the way how these two situations (repeated extreme abduction and/or long-term moderate abduction) get incorporated into the working practices of a large number of people (hundreds of thousands in shoemaking in Mexico, for example). Besides, chapter 2 included a section (2.8) reviewing the effects that those working conditions have been shown to exert on the shoulder and (albeit to a lesser extent) the neck, which has earned them a place as one of the most significant ergonomic stressors. It is in view of these facts that it was decided for the present investigation to look into postures which create loads on the muscles of the shoulder by holding the arms abducted and unsupported.

Even though, as has been previously mentioned, the anatomical structures of the shoulder region come under excessive stress both when the arm is abducted to more than 90° and when it is subjected to long-term abduction at lower angles, it was decided that the postures studied would involve the abduction of both arms at angles of 30°, 60°, and 90°. This decision was taken because, without demeaning the importance of the stress created by extreme abduction and its undesirable effects, it is the experience of this researcher that the situation where the arms are kept moderately abducted for a long time is found much more frequently, involving larger numbers of workers. Therefore, if the abduction angle was to be kept to a maximum of 90°, it was assumed that an interval of 30° between pairs of postures would be enough to create significant differences in the extent of both the physiological changes brought about by fatigue and the subjective perception of those changes.

Thus, the subjects who participated in the first experimental stage of the study were required to hold only one of such postures, with the arms abducted at 60°. The second series of experiments had the subjects holding the three postures, with arms abducted at 30°, 60°, and 90°. These three postures are illustrated in figure 3.1 a), b) and c), respectively. In all the cases the subject held the posture whilst standing upright without wearing shoes.

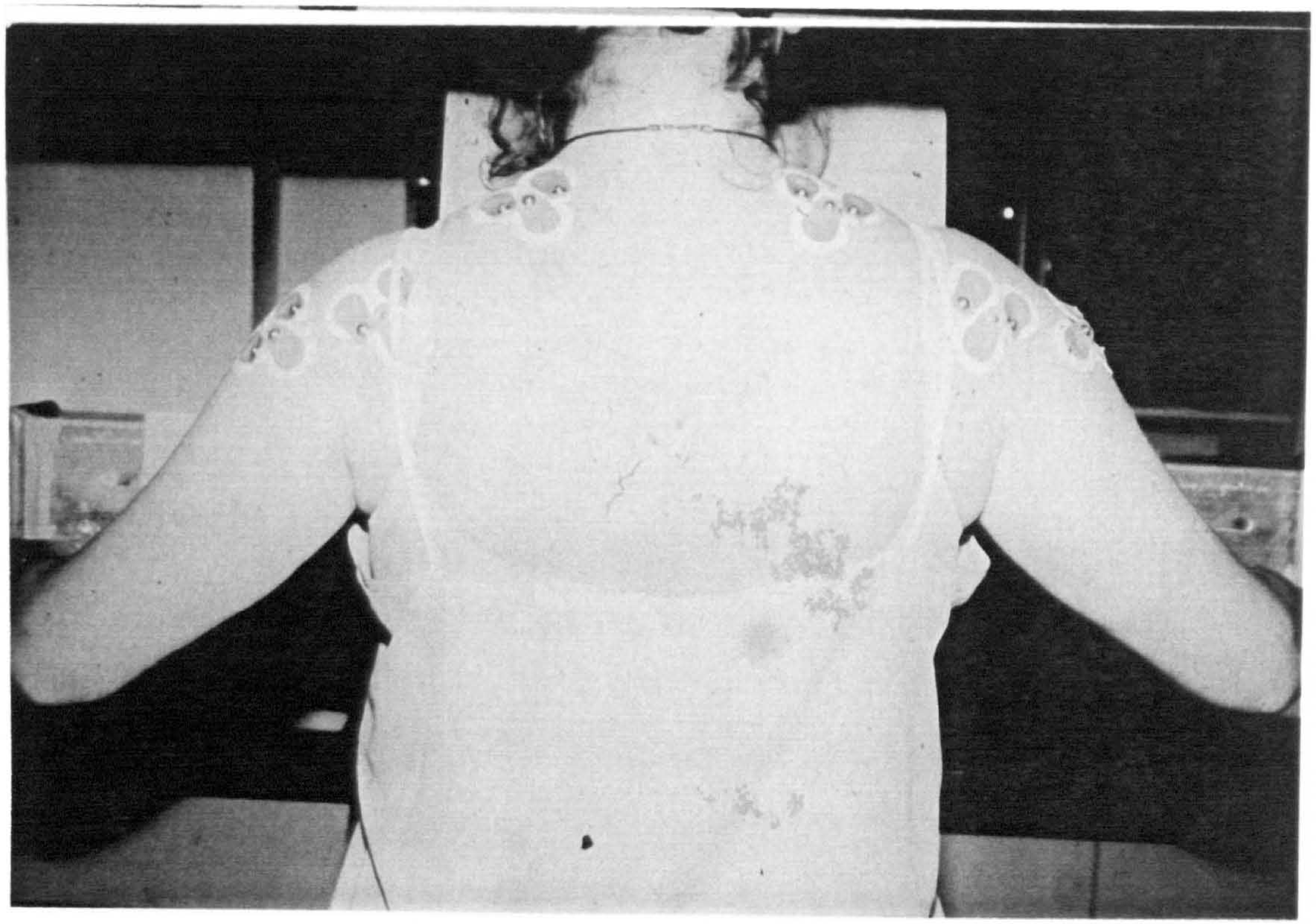


Figure 3.1 a) Rear view of a subject holding the posture with arms abducted at 30°.

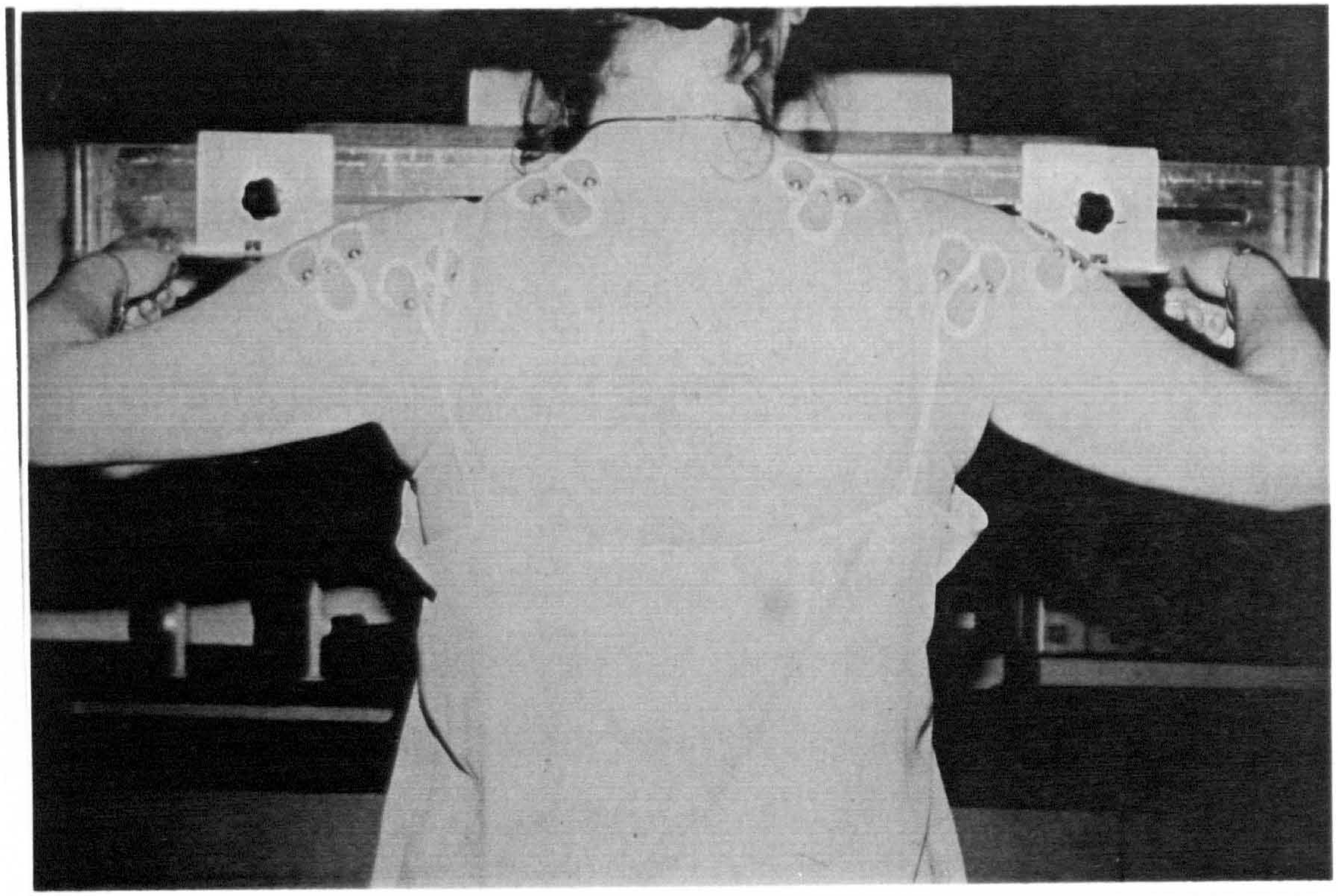


Figure 3.1 b) Rear view of a subject holding the posture with arms abducted at 60°.

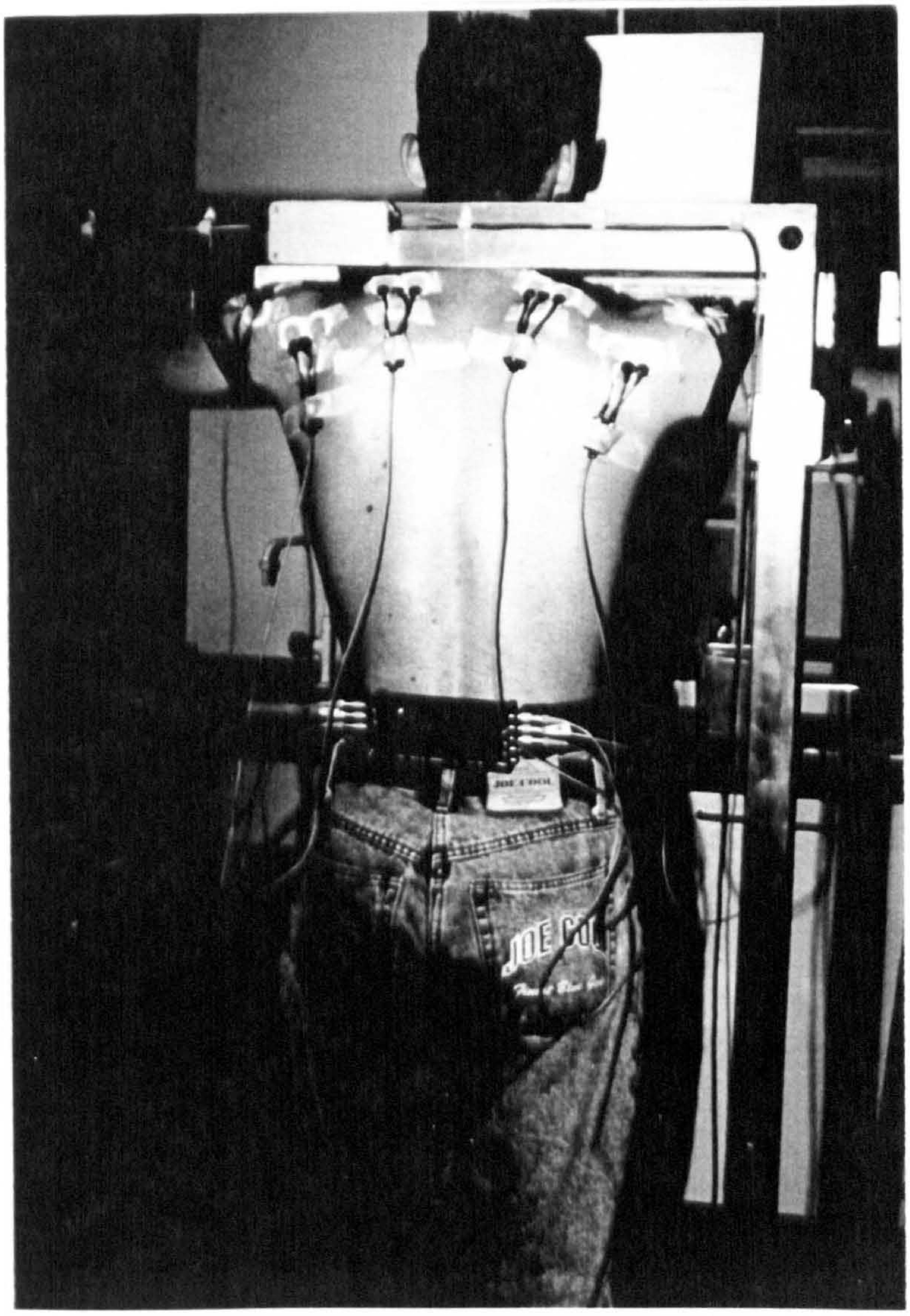


Figure 3.1 c) Rear view of a subject holding the posture with arms abducted at 90°.

The experimental postures required the subject to place the upper arms in the coronal plane, forming with the trunk the abduction angle of interest. The elbows were flexed at 90° and the forearms were kept horizontal. The wrists were held straight and the palms of the hands faced each other. During the holding of the postures the experimenter checked frequently the body segments to ensure that they were kept in the correct position.

3.4.3 Layout of the experimental chamber

All the experimental work was conducted in a single setting, in an area part of a large laboratory with adequate illumination and ventilation. The climatic conditions inside the laboratory could not be controlled. However, the room temperature did not show considerable variations throughout the period when the experiments were conducted, it remained around 20 to 22 °C.

The experimental apparatus was purpose-built. Figures 3.2 and 3.3 show diagrams of its side and front views, respectively. The apparatus was based on a rectangular steel platform (labelled A in figure 3.2) on which the subjects stood during the holding of the postures. In the middle of one of the narrow ends of this platform, screwed onto it, there was a cylindrical pole, labelled B in figure 3.2. Attached to the pole by means of a ring that could slide up and down it there was a rectangular wooden platform (labelled C in figure 3.2).

A piece of cardboard, with a body diagram and a scale for the rating of subjective perception stapled onto it, was placed on top of this wooden platform during the trials of posture holding. The height of the platform was set by the height of the subject's hands in the posture being held. Attached to the end of the wooden platform that faced towards the subject came a vertical metallic plate, which is labelled D in figure 3.3. This had two symmetrical slots, along which two small L-shaped metallic pieces (labelled E in figures 3.2 and 3.3) could slide to left and right. The position of these pieces on the vertical plate was determined by the distance between the hands of the subject. Each of this sliding pieces had attached to it a small rectangle of a very light plastic material. To help them keep their hands in the position required by the experimental posture, the subjects were asked to pinch very lightly on this piece of plastic with their thumb and forefinger. Because of its size and its very small weight, this marker of the hands' position could not offer the subject any support at all.

Figure 3.4 shows the way in which the wooden platform and the ancillary pieces of equipment were arranged in order to help the subject to remain in the required posture (abduction at 30° in the instance illustrated).

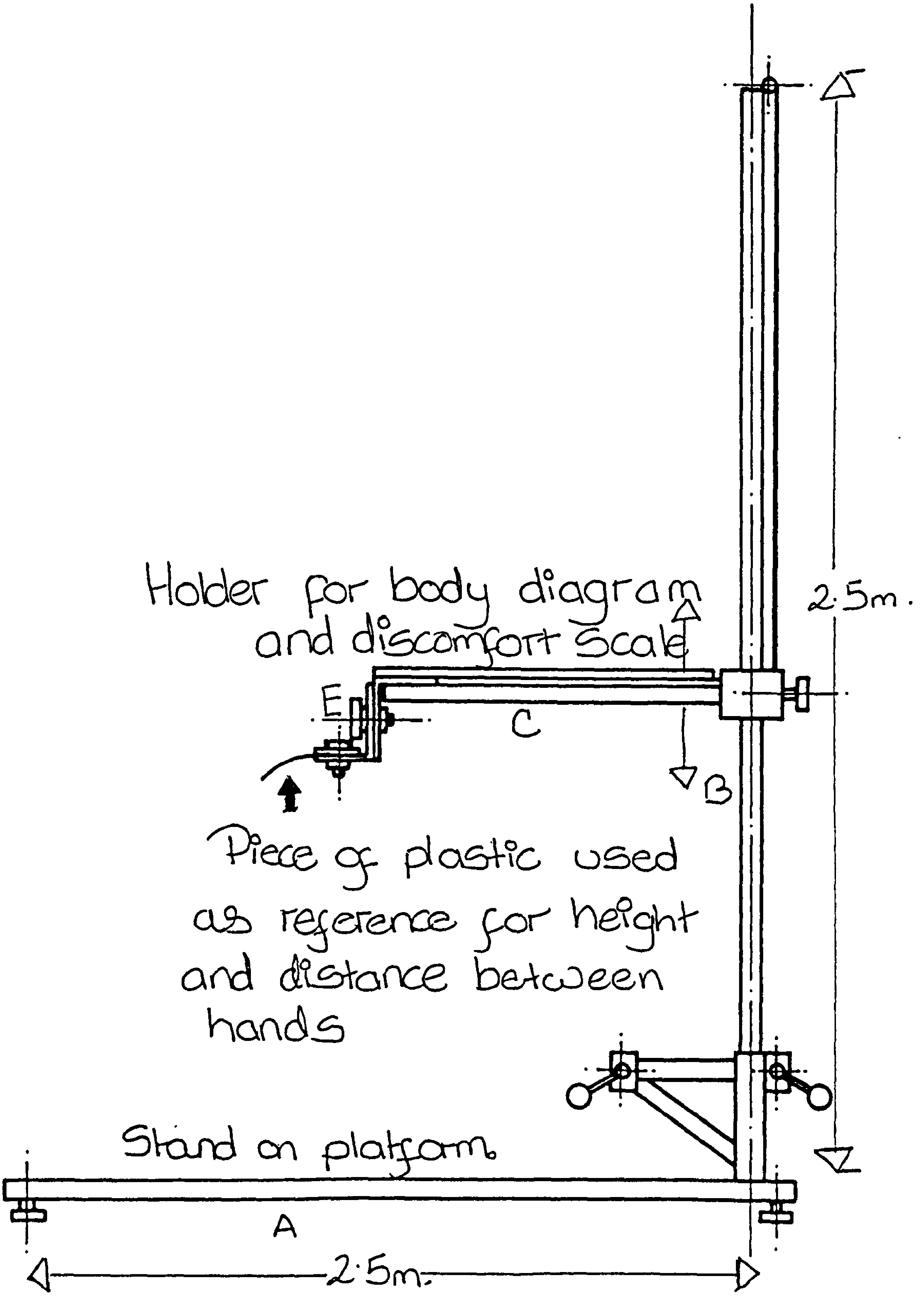


Figure 3.2 Diagrammatic side view of the experimental apparatus.

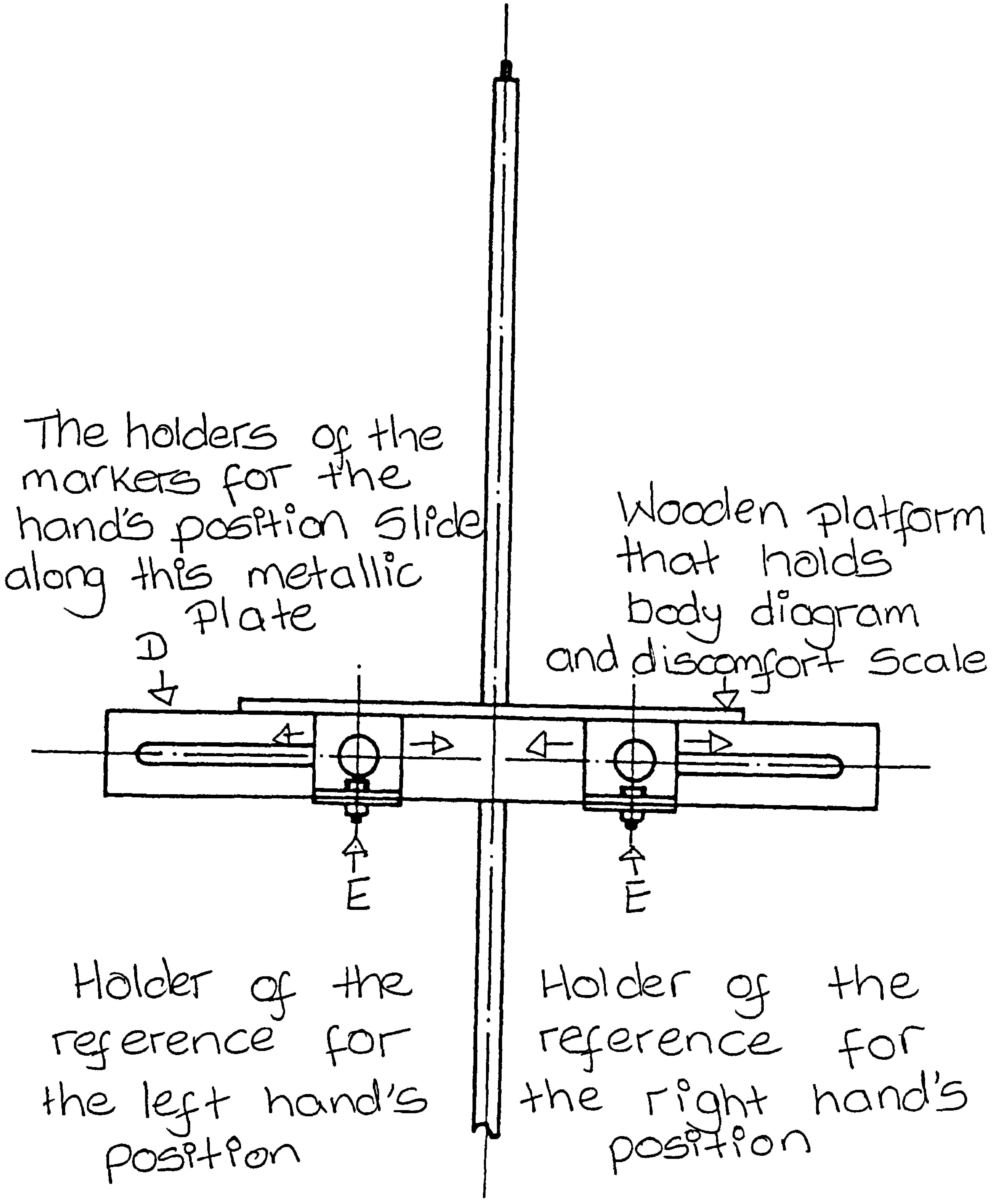


Figure 3.3 Diagrammatic front view of the experimental apparatus.

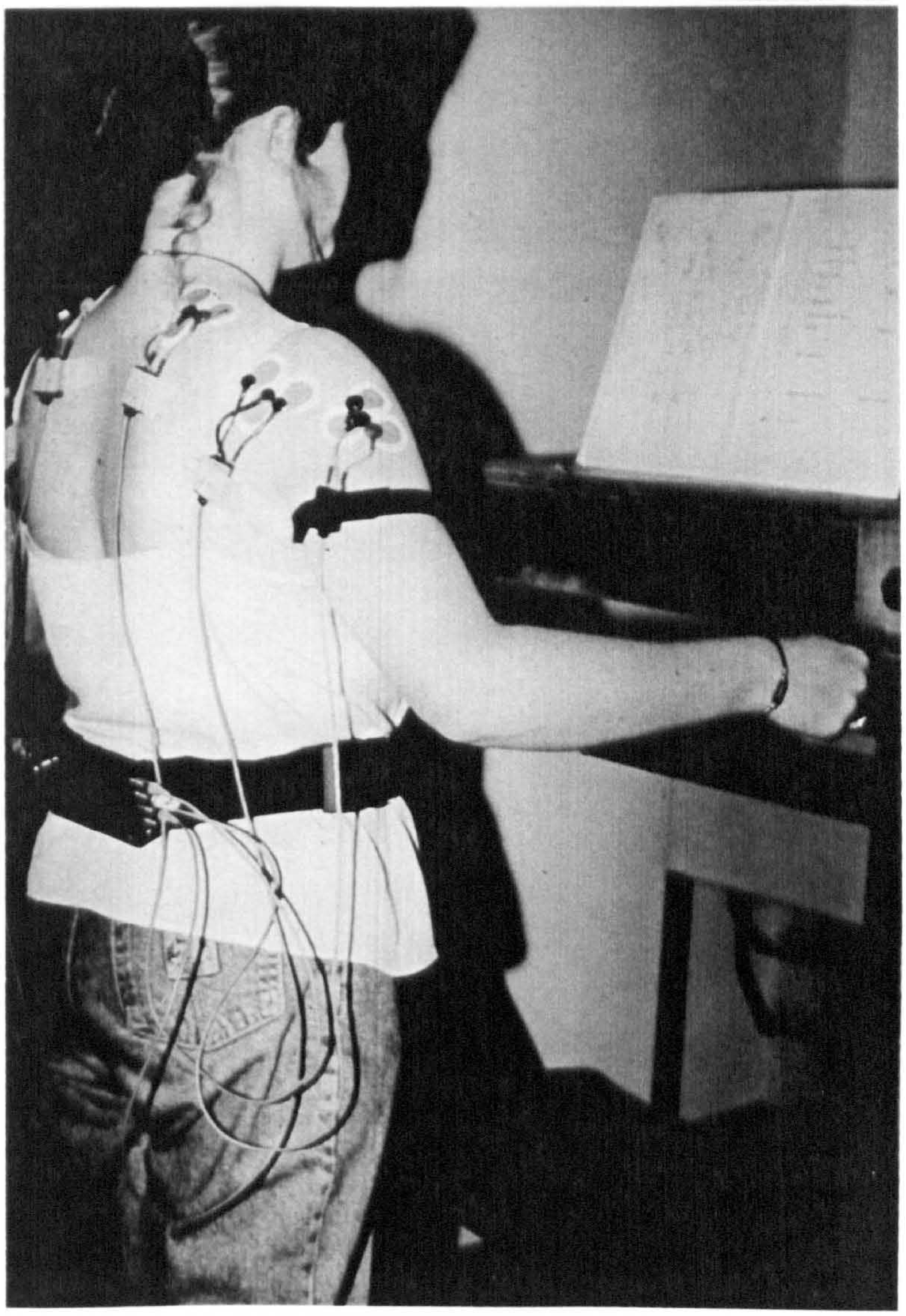


Figure 3.4 A close view of the experimental set-up employed to help the subject to remain in the required posture.

3.4.4 Preliminary procedures

In both experimental stages the first visit of the subjects to the laboratory was used for preliminary preparations. The first of these was the measurement and recording of the relevant anthropometric data: weight, stature, shoulder height, arm length and forearm length. On arrival to the laboratory, the subjects were asked to remove their shoes and step onto a stadiometer, for the weighing and measurement of stature. After this, their shoulder height was measured using a scale fitted to the stadiometer. Next, arm and forearm lengths were measured with an anthropometer (Holtain, U.K.). These body dimensions were measured following the procedures laid down by Pheasant (1986).

After completing the anthropometric measurements the subjects moved into the experimental chamber. There, they were placed by the researcher in each of the three postures studied and the rig was adjusted accordingly, in order to ensure that the postures could be exactly replicated once the trials got under way.

To get the subject into the experimental postures, they started standing (shoeless) on the steel platform (A) and facing the wooden platform (C) mounted on the pole (B). The subject was asked to keep the arms hanging by the sides in a natural manner. The researcher then placed a pendulum goniometer (MeDesign, Ltd, U.K.) on the mid-point of the lateral aspect of the right upper arm, zeroed it and then asked the subject to slowly raise the arm to the side, keeping it stretched and aligned with the body, until the goniometer indicated that the arm had reached the abduction angle desired (30, 60 or 90°). Keeping the arm at that angle the subject then bent the elbow to 90°, and this angle was checked by the researcher using an universal goniometer. Next, the subject was asked to rotate the forearm 90°, such that its internal aspect and the palm of the hand turned towards the body's medial line. Whilst the subject held the arm in that position, the researcher slid the wooden platform up or down the pole until the piece of light plastic mounted on the L-shaped piece (E) was roughly at the same height as the index finger. Then, the researcher slid the L-shaped piece to the right or left as needed, so to bring the piece of plastic near the index finger. At this stage, the subject's right arm was in the required position. After completion of these initial adjustments, the subject put the right arm down and the same procedure was followed to place the left arm in the required position.

Having positioned right and left arm separately, the researcher then asked the subject to raise both arms until their hands were roughly at the same height as the wooden platform. Once the subject was in that posture, the height of the platform and the position of the L-shaped pieces were carefully adjusted, so as to bring the outer corner of the plastic piece to rest exactly between thumb and index of the corresponding hand. Keeping the subjects as they were at that point, the researcher checked again that the upper arms were in line with the subject's trunk, the shoulders were abducted at the angle desired, the elbows were flexed at 90°, the wrists were straight and the palms of the hands faced each other. Once the researcher felt satisfied that the subject had reached the experimental position, to mark the exact location of the subject's feet, a steel bar was placed on the stand-on platform and dragged towards the subject's feet until both first toes were in contact with the bar. The subject was then asked to clear the experimental chamber and the height of the wooden platform, the distance of the L-shaped pieces from the centre of the metallic plate where they were mounted, and the distance of the foot-marker from the edge of the platform were recorded, so that in subsequent sessions they could be placed in exactly the same position, so ensuring that the subject always adopted the same posture. Figure 3.5 shows the whole set-up of the experimental chamber, as arranged for one of the female subjects who participated in the second experimental stage. On that occasion the subject was holding the arms abducted at 30°.

3.4.5 Collection of discomfort ratings

In both experimental stages, the trials consisted basically of asking the subject to remain in a fixed posture until reaching their limit of endurance. This limit had already been defined to the subject during the recruitment phase, when it was described to them as the most unpleasant sensation of muscular discomfort they could possibly bear, regardless of where in the body it was located. At that stage, the researcher took special care in discussing with each subject how they perceived such limit, so that the concept of 'unbearably unpleasant' was, as much as possible, agreed between them.

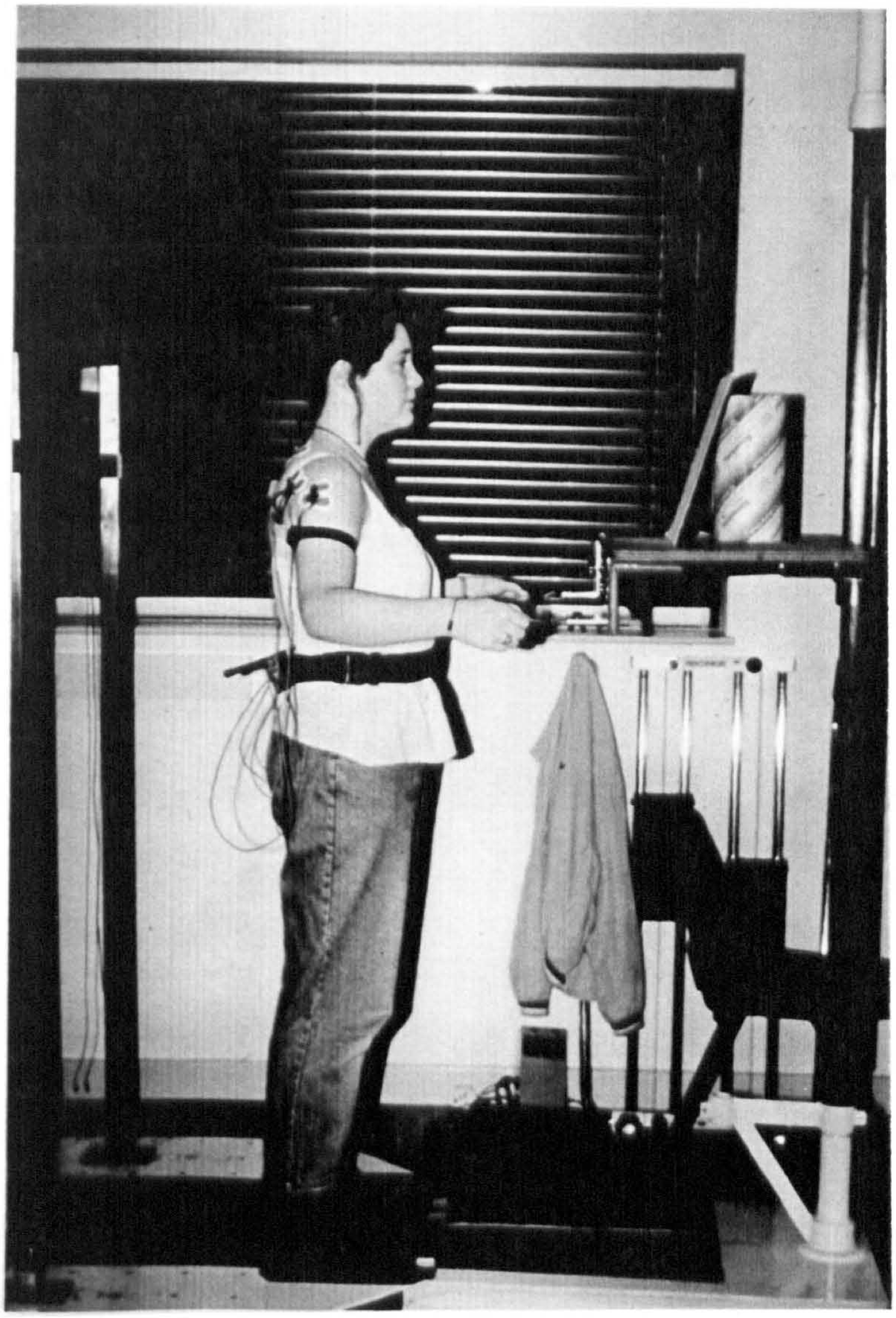


Figure 3.5 Full view of the experimental set-up used during the measurement of maximal endurance to arm abduction. The case illustrated was the holding at 30°.

The endurance limit has been defined in a similar way in virtually every study with an interest on the subjective perception of the changes provoked by muscular exertion, regardless of the mode of such exertion. In the studies by (among many others) Barbonis (1979), Hagberg (1981a), Milner (1985), Manenica (1986), van der Grinten (1991) the interest was centred on the growth of discomfort. Pain has been the focus of interest in many other studies, like those by Caldwell and Smith (1966), Kirk and Sadoyama (1973), Kilbom et al (1983). The rating of the perceived effort has also been used as the criterion to set the limit to muscular exertion, as in the studies by Kilbom et al (1983), Rohmert et al (1986), Hasson et al (1989). Thus, the definition of endurance limit as applied in the present study is a concept widely accepted .

Since one of the aims of the study was to determine how the discomfort changed during the holding of the posture, the subjects had to provide a sequence of discomfort ratings, from the beginning of the trial right to its end. Thus, obtaining information on the magnitude of the discomfort experienced by the subject was a very important aspect of the experimental work, not least because in the end it would permit to know which body parts bore the most of the postural stress.

The tools used to collect the discomfort ratings were a body map, a modified version of the one used by Corlett and Bishop (1976), and a rating scale, developed by Borg (1982). Figure 3.6 shows the body map used, and the scale for the rating of discomfort is reproduced in figure 3.7. During the trials, these two instruments were in full view of the subject, attached to the piece of cardboard that was placed on the wooden platform, directly in front of them. This arrangement may be seen in figure 3.4.

The reason for the use of Borg's category-ratio scale is threefold. First, as Borg (1982) pointed out, the incorporation into a single instrument of category and ratio properties adds the precision imparted by a ratio scale (in which the interval between two adjacent marks has the same relative value), to the ease of use of a category scale. Second, it has been shown to be a valid and accurate method for the rating of subjective perceptions of discomfort, pain (Harms-Ringdahl et al 1983), fatigue or degree of exertion (Rohmert et al 1986, Jørgensen et al 1988, Hasson et al 1989) and applied in studies involving both static and dynamic exercise. Third, since the use of Borg's category-ratio scale is becoming widespread, reporting the discomfort ratings on this scale will make it possible for the results of the present investigation to be related to those obtained from a wide variety of applications.

The duration of the trials was registered using stopwatch. This was started as soon as the experimenter was satisfied that the subject had adopted the required experimental posture. To collect the discomfort ratings, every 60 seconds (120 seconds at some stages in the larger trials) the researcher called out in alphabetic order the letters that identified the body regions, and the subject had to respond to each prompting with a value taken from the scale.

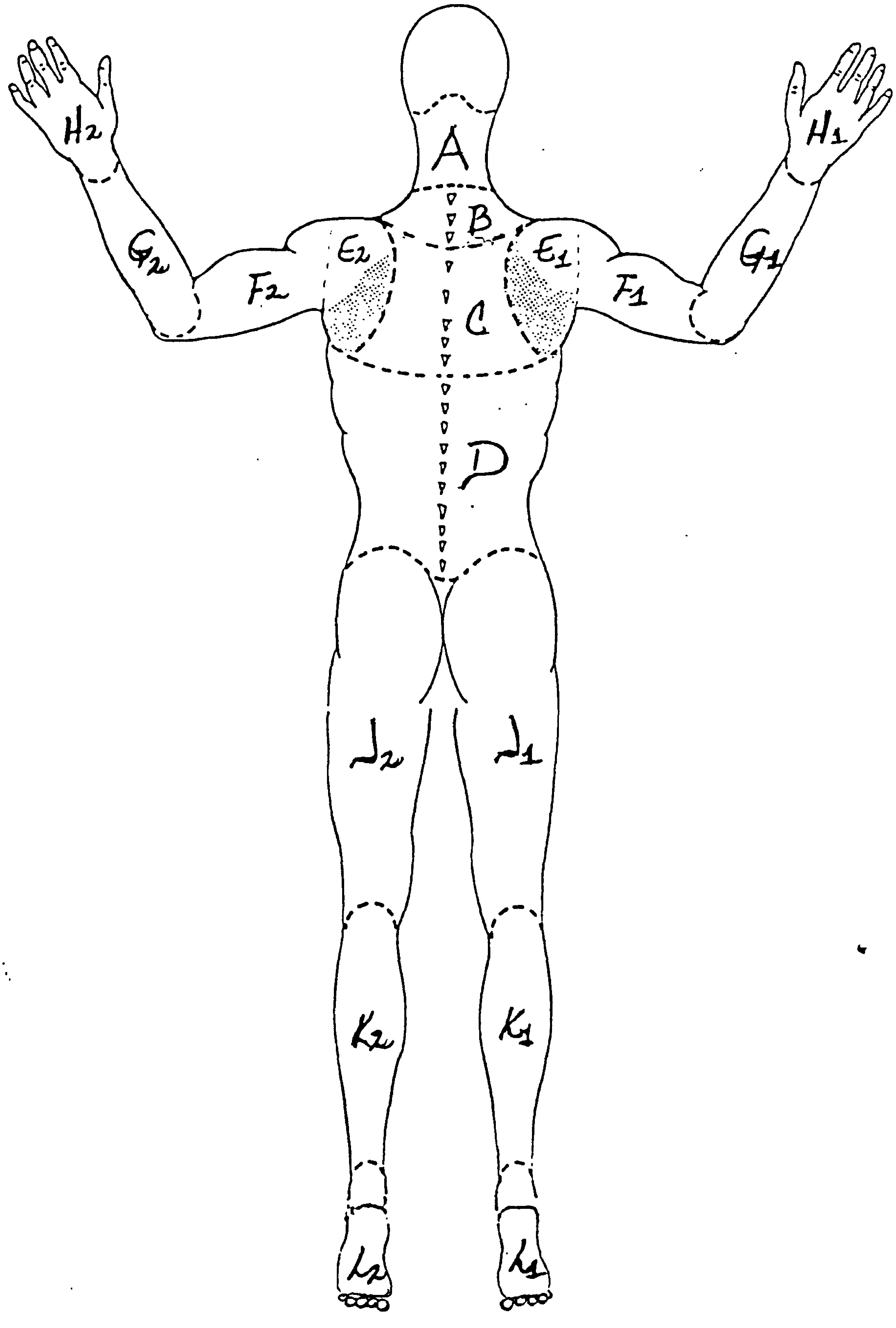


Figure 3.6 Body map used for the collection of discomfort ratings. Modified from Corlett and Bishop (1976).

0	Nothing at all	
0.5	Extremely weak	(just noticeable)
1	Very weak	
2	Weak	(light)
3	Moderate	
4	Somewhat strong	
5	Strong	(heavy)
6		
7	Very strong	
8		
9		
10	Extremely strong	(almost max)
●	Maximal	

Figure 3.7. Category-ratio scale for the rating of discomfort. Reproduced from Borg, 1982.

Although they were marked as such on the body diagram, the regions E to L were not called out separately for left and right (by number 1 or 2), but it was left to the subject to report them in that way, when they judged that the discomfort was not the same on both sides of the body. The end point of the trial was determined by the subjects themselves, who called for the holding to stop because the discomfort had got unbearable, meaning that they had reached the limit of their endurance to the posture.

Even though the experimental postures were designed expressly to place the strain mainly on the shoulder (and probably the neck), it was deemed necessary to offer the subjects a diagram that included the whole body rather than just those two regions, considering the possibility that they could hold the postures for long enough to experience considerable discomfort in some other part of the body, and this was information that obviously should not be missed.

3.4.6 Procedures Followed in the Collection of EMG Signals

The electromyographical signals were collected from the descending portion of the trapezius muscle and from the medial and the posterior portions of the deltoid muscle, on both arms. The reasons for this choice of muscles, the procedures followed in the analysis and interpretation of the signals, and the results from that analysis will be presented in chapter 7. However, the methods applied in the collection and conditioning of EMG signals will be described at this point.

The equipment used, and the procedures applied for the collection, analysis and interpretation of the electromyographic information fully complied with the recommendations issued by the International Society of Electrophysiological Kinesiology (Winter et al, 1980). The signals were collected and conditioned using biotelemetry equipment (MT8-3 Radio Telemetry System, MIE Medical Research, Leeds, U.K.). It consisted of preamplifiers (8k, CMRR = -114 dB), transmitter unit and receiver unit, with integrated skin resistance meter.

The electromyographic signals were picked up by disposable Ag-AgCl electrodes (P-00-S type, Medicotest A/S, Ølstykke, Denmark) arranged in bipolar configuration. The distance between the pick-up areas was 30 mm, with the collecting electrodes placed parallel to the muscle fibre. A third electrode was used as a local reference, and a metallic plate was placed on the subject's right wrist, to act as reference for the whole arrangement. Although the electrodes were pre-gelled, the additional application of an electrolytic paste (Clinical Products, Rome, Italy) helped to achieve the best conduction of the signal.

To locate the site of application of the electrodes on the medial deltoid, the subjects were asked to place their arm by their side, with the elbow flexed at 90°, and attempt an abduction against the resistance applied by the researcher with one of his hands, whilst with the other he felt the subject's arm for the belly of the muscle. A similar procedure was used for the posterior deltoid, only in this case the subject attempted to move their arm backwards. On the trapezius, the electrodes were placed so that the mid-point between them was located at approximately 5 cm from the 7th cervical vertebra, on a line joining that bone to the acromion.

The preparation of the skin in the sites where the signal was collected involved shaving (when necessary), cleaning with alcohol and, to lower the resistance, application of an abrasive paste (Omni-prep™, D.O. Weaver & Co., Aurora, CO, USA). Once the skin was prepared, the electrodes were applied and the resistance checked with the skin resistance meter built into the receiver unit of the telemetry equipment. If the reading was above 5 kΩ, the skin was cleaned again and more abrasive paste applied. Typically, the skin resistance could be lowered to between 0 and 2 kΩ.

Once the electrodes were in place, they were connected to the preamplifiers (located at a distance of approximately 8 cm). Here, besides being amplified, the EMG signal was filtered (pass band 0-165 Hz). The signals were then relayed to the transmitter unit and from this to the receiver unit, where they were further filtered. From the receiver unit, the data were continuously relayed in real time onto a personal computer using the 12-bit data acquisition card and PROCURE software supplied with the MT8-3 telemetry system. The signals were presented on the computer's monitor, allowing a continuous check for the existence of any obvious irregularity. The next stage, spectral analysis of the EMG signals, will be described in chapter 7.

CHAPTER 4

TESTING OF MILNER'S MODEL FOR THE PREDICTION OF RECOVERY FROM POSTURAL EXERTION

4.1 Introduction

It was mentioned in chapter 2 (section 2.11.2) that Milner (1985) developed a model that can be used to predict the proportion of endurance to postural load that will remain after a person has rested following a period of postural exertion. The model (Milner's model) is expressed by the equation

$$\text{Recovery} = (\text{MHT}-\text{HT}) + \text{HT}[e^{-0.164(\text{HT}/\text{Rest})}]$$

where MHT is the maximum holding time, HT is the length of time that the posture is held, Rest is the length of time allowed for relaxation following the holding and Recovery is the remaining endurance, which may be expressed either as a length of time or as a percentage of the maximum holding time (depending on the units in which the other parameters are expressed).

This model appeared as a promising starting point for the present investigation. The logical progression should be to carry on from the point where Milner left: to increase the number of periods of posture holding and rest for, even though Milner attempted to extend the model to situations of multiple combinations of postural exertion and rest, it actually worked successfully only for the instance of a single combination of posture holding and rest. Besides, going into multiple combinations work-rest would mean moving closer to the simulation of a real-life situation of intermittent work, thus allowing to test whether the model would still work under those conditions.

Before that, however, two crucial questions had to be answered. First, will the model still hold when the posture studied is different from that which was used to develop it? This was a posture where the person bent forwards and worked with the arms fully stretched to the front, so creating loads mainly on the muscles of the low back and legs. However, the interest of the present research was on postures where the arms were abducted in the coronal plane, creating loads around the shoulder joint. The second question is whether the model's predictions will extend to female subjects.

The decision about the postures to be studied was reached at a very early stage in the planning of the present investigation. The reasons behind such decision appeared self-evident to this researcher, and have already been discussed in the previous chapter. However, the decision in relation to the gender of the subjects was taken much later, when the review of the relevant literature showed that an overwhelming majority of the studies in the field of isometric exertion (particularly those conducted in a laboratory environment) have been carried out on male subjects and seldom, if ever, is this choice of subjects discussed. To this researcher it looked as if this was rather a case of 'going with the flow', or perhaps it might have to do with the availability (or willingness to endure the rigours entailed by the experimental procedures) of the potential subjects, which might be assumed to be larger for male subjects. The study by Milner (1985) was no exception to this trend, as neither was that by Barbonis (1979), from which the former took its cue.

Therefore, it was a spell of sheer curiosity which, besides the obvious (and highly significant, no doubt) question regarding the effect that the change of posture could have on the performance of the model, suggested a second question: will Milner's model hold its validity when it is translated across the gender divide? This chapter deals with the experimental work performed in the search of the answer to those questions and the results obtained from it.

4.2 Procedures

Since the ultimate aim of this experiment was to put to the test Milner's model, the experimental arrangements needed to replicate as closely as possible those applied in the development of the model, with the exception of the posture studied and the gender of the subjects.

The procedures and criteria followed in selecting the subjects have been presented in the previous chapter. Eight subjects took part in this first experimental stage. This was the number necessary to recreate the experimental design applied by Milner to develop the model. They attended the laboratory on a total of ten occasions, separated from each other by a minimum of 2 days. The first session was used to carry out the anthropometric measurements on the subject and to adjust the experimental rig to each individual's dimensions, in the way already described in chapter 3. The anthropometric details of the subjects are presented in table 4.1. The second session was devoted exclusively to the assessment of the maximum endurance to the posture, by measuring the longest time the subject could hold it. Each of the eight remaining sessions replicated one of the experimental conditions used by Milner in the development of the model.

4.2.1 Determination of the sites of maximum discomfort

As mentioned in chapter 3, as part of the recruitment process, the subjects were given detailed information about the procedures involved in the experimental work, including the collection of discomfort ratings. However, because this information was quite important for the subsequent stages of the investigation, the procedure was again explained when they came to the laboratory to be measured up, and this time the way to identify the limits between the regions depicted in the body diagram (shown in figure 3.6) was also explained. The researcher also emphasised that each rating they returned should reflect the sensations they were experiencing at that very moment, and that they should not try to remember the ratings they had returned before. However, they were also told they were expected to be capable of identifying and remembering as clearly as possible the sensation they would rate as 'unbearable discomfort' since, for the forthcoming sessions, only when they experienced that very sensation they should call for the effort to stop.

Table 4.1 Anthropometric characteristics of the subjects who took part in the trials to test Milner's model of recovery from postural exertion.

Subject's identifier	Age (years)	Weight (kilograms)	Stature (mm)	Shoulder height (mm)	Arm length (mm)	Forearm length (mm)
No. 1	20	56.1	1763	1445	743	501
No. 2	21	58.7	1676	1368	708	476
No. 3	22	57.1	1600	1321	676	459
No. 4	21	71.6	1733	1425	725	495
No. 5	20	57.2	1672	1386	700	475
No. 6	19	60.8	1603	1277	675	454
No. 7	20	57.5	1653	1344	698	470
No. 8	20	53.3	1578	1288	660	448
Mean ± s.d.	20.4±0.92	59.04 ± 5.504	1660 ± 65	1357 ±61	698 ±28	472 ± 19

During the trials, the first value of discomfort rating was always obtained a few seconds before the subject was asked to adopt the experimental posture; the subsequent ratings were collected every 60 seconds (120 seconds at some stages of the larger trials) throughout the holding of the posture, except for the last one that was obtained at the moment the subject declared their intention of stopping the effort because of the unbearable discomfort. As soon as this happened, they were asked to mark on a copy of the body diagram where had they just experienced that sensation. This information was crucial to determine what body parts were being subjected to the major efforts during the holding of the posture, and so defining the most likely locations from where to obtain the electromyographic information during the second stage of the investigation.

4.2.2 Measurement of the maximum holding time

The second experimental session was used exclusively to measure the subject's maximum holding time. This was recorded as the time elapsed between the moment when the subject adopted the experimental posture and the moment when they informed the researcher that the discomfort had got unbearable and they were about to abandon the exertion. On this basis, the maximum holding time will be assumed to give the measure of the subject's endurance to the efforts created by the holding of the posture.

Before starting the stopwatch used for recording the holding time, the researcher checked that the shoulders, the elbows and the wrists of the subject were placed at the correct angles. The checking was repeated frequently (several times every minute, at some stage) during the holding and if any of the angles had changed by more than 5 degrees in any direction, the subject was instructed to correct the deviation and get back to the posture. These checks were also carried out in all of the subsequent sessions.

4.3 Experimental design

The remaining eight sessions were used to replicate the experiments performed by Milner to develop the model. These consisted in the execution of a sequence posture holding /rest /posture holding. The length of the first period of posture holding and of the rest period were set as a proportion of the maximum holding time already known. The second holding of the posture was intended as the measurement of the endurance capacity that remained following the initial effort and the rest, and was therefore sustained by the subject until they reached the point of unbearable discomfort. From now on, this second holding will be referred to as 'holding to exhaustion', to distinguish it from the holding performed in the second experimental session which, as previously mentioned, was taken as the measure of the absolute endurance.

Eight combinations of initial holding and rest times were used to develop the model and were therefore replicated in this series of trials. Each combination was identified by a capital letter, as shown in table 4.2.

Table 4.2 Composition of the combinations of work and rest performed during the testing of Milner's model.

Combination	Length of the initial holding (as % of the maximum holding time)	Length of the rest period (as % of the maximum holding time)
A	25	25
B	25	12.5
C	50	50
D	50	25
E	75	75
F	75	37.5
G	100	100
H	100	50

The combinations were designed in pairs (A-B, C-D, E-F, G-H), such that each length of initial holding was followed by a period of rest which was either as long as that holding or half its length.

As previously mentioned, the length of the initial holding and the rest were calculated as a function of the maximum holding time measured during the second session. However, that convention could not be applied to combinations G and H. On those occasions the subject was required to perform an initial holding that should be 100% of the maximum holding time (which in fact made them a second and third measurement of the maximum endurance to the posture, respectively). In consequence, the length of the rest for combinations G and H could not be calculated as a proportion of the original maximum holding time, it had to be referred to the length that the initial holding reached on each occasion. Accordingly, the remaining endurance was also calculated as a proportion of this initial holding.

The eight combinations of posture holding and rest were assigned to each subject according to the 8x8 randomised latin square shown in table 4.3. This approach was taken to ensure that, by presenting the conditions in a different order to each subject, this would balance any possible training effect that could develop as they completed the successive trials.

Table 4.3 Order in which each subject performed the trials combining holding time, rest and holding time to the endurance limit, with arms abducted at 60°. The composition of the combinations is shown in table 4.2.

	Order of presentation							
Subject	1st	2nd	3rd	4th	5th	6th	7th	8th
No.1	C	D	G[1]	H[2]	F	E	A	B
No. 2	B	H[2]	D	G[1]	E	A	F	C
No. 3	G[1]	F	H[2]	D	B	C	E	A
No. 4	E	B	A	F	D	H[2]	C	G[1]
No. 5	D	C	F	A	H[2]	G[1]	B	E
No. 6	A	E	B	C	G[1]	E	H[2]	D
No. 7	H[2]	A	E	B	C	D	G[1]	F
No. 8	F	G[1]	C	E	A	B	D	H[2]

[1] Second maximum holding time. [2] Third maximum holding time

4.4 Results

4.4.1 Maximum holding time and degree of recovery

Table 4.4 presents the length of the maximum holding time achieved by each subject during the measurement performed in the second session. Since the length of the initial holding and rest for the eight remaining sessions presented in table 4.2 (and shown next to the session's identifier in table 4.4) were given only as proportions of the maximum holding time, they are also included in table 4.4, now expressed in actual units of time. Whilst the length of the elements in combinations A to F was calculated on the basis of the maximum holding time shown in the second column of table 4.4, that of the elements in combinations G and H did not depend on the latter, rather they stand on their own.

All the times presented in table 4.4 and subsequent ones are given in metric minutes, to allow for comparison with the results obtained by Milner, who used such units to measure the length of the holding and rest periods, as well as that of the remaining endurance (or, in Milner's words, recovery).

Subject No. 6 performed only four of her combinations. She developed a muscular trouble in the neck (in circumstances not related to the trials) and the researcher decided to release her from taking part in any further sessions, in order to avoid worsening her condition.

Table 4.4 Maximum holding time and length of the combinations holding/rest for the trials with arms' abduction at 60°. The times are expressed in metric minutes.

Combination performed									
Subject	Maximum holding time (minutes)	A (25/25)	B (25/12.5)	C (50/50)	D (50/25)	E (75/75)	F (75/37.5)	G[1] (100/100)	H[2] (100/50)
No. 1	14.8	3.7/3.7	3.7/1.8	7.5/7.5	7.5/3.8	11.0/11.0	11.0/5.5	13.7/13.7	12.0/6.0
No. 2	15.5	4.0/4.0	4.0/2.0	7.8/7.8	7.8/3.9	11.7/11.7	11.7/5.8	15.4/15.4	15.4/7.7
No. 3	12.8	3.2/3.2	3.2/1.6	6.3/6.3	6.5/3.2	9.6/9.6	9.5/4.8	12.8/12.8	12.8/6.4
No. 4	13.3	3.3/3.3	3.3/1.7	6.7/6.7	6.7/3.3	10.0/10.0	10.0/5.0	11.3/11.3	13.3/6.7
No. 5	9.4	2.4/2.4	2.3/1.2	4.7/4.7	4.7/2.3	7.0/7.0	7.0/3.5	7.9/7.9	8.8/4.4
No. 6	6.7	1.6/1.6	1.7/0.8	3.3/3.3	-----	5.0/5.0	-----	-------	-------
No. 7	7.7	2.0/2.0	2.0/1.0	4.1/4.1	4.0/2.0	5.8/5.8	5.8/2.9	7.4/7.4	7.4/3.7
No. 8	13.8	3.5/3.5	3.5/1.8	7.4/7.4	7.1/3.5	10.4/10.4	10.4/5.2	13.5/13.5	8.7/4.3

[1] Second maximum holding time. [2] Third maximum holding time.

Table 4.5 presents the length of the holding to exhaustion that followed the combination of initial holding and rest in each of the eight sessions. This value measures the remaining proportion of the maximum endurance to the posture (or recovery) and it is shown as actual time (in metric minutes) and as percentage of the subject's maximum holding time, in order to allow the comparison (to be done in the discussion section) with the values predicted by Milner's model, which was said by its author to be suitable for use with either type of units.

The extent of the recovery predicted by Milner's model for each of the combinations of initial holding and rest is shown in table 4.6. Similarly to the observed recovery, the predicted recovery is presented as the actual length of time in metric minutes and as percentage of the subject's maximum endurance.

Table 4.5 Length of the holding to exhaustion that followed the combinations of initial holding and rest, in the trials with arm abduction at 60°. The values are shown both in metric minutes and as proportion of the subject's maximum holding time.

Combination	A		B		C		D		E		F		G		H	
Remaining endurance	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes
Subject No. 1	94	13.9	72	10.6	52	7.7	52	7.7	66	9.7	57	8.4	54	7.4	81	9.7
Subject No. 2	89	13.8	58	9.0	81	12.6	77	11.9	60	9.3	66	10.3	74	11.4	44	6.8
Subject No. 3	98	12.5	105	13.5	84	10.7	80	10.3	77	9.9	96	12.3	63	8.0	75	9.6
Subject No. 4	83	11.0	70	9.3	86	11.5	87	11.6	50	6.6	60	8.0	82	9.3	53	7.0
Subject No. 5	93	8.7	83	7.8	94	8.8	71	6.7	80	7.5	67	6.3	75	5.9	76	6.7
Subject No. 6	110	7.4	109	7.3	87	5.8	---	---	84	5.6	---	---	---	---	---	---
Subject No. 7	83	6.4	75	5.8	78	6.0	95	7.3	77	5.9	82	6.3	72	5.3	72	5.3
Subject No. 8	70	9.7	54	7.5	71	9.8	63	8.7	65	8.5	79	10.9	59	7.9	76	6.6
Mean ± s.d.	90.0 ±11.9	10.42 ±2.85	78.2 ±20.0	8.85 ±2.377	79.1 ±12.88	9.11 ±2.496	75.0 ±14.5	9.17 ±2.106	69.9 ±11.56	7.87 ±1.719	72.4 ±13.87	8.93 ±2.313	68.4 ±9.98	8.93 ±2.313	68.1 ±13.92	7.39 ±1.643

Table 4.6 Proportion of the maximum endurance to postural loading that Milner's model predicted should remain following the combinations of initial holding and rest, in trials with arm abduction at 60°. Values shown as actual time (in metric minutes) and as percentage of the maximum endurance (%MHT).

Combination	A		B		C		D		E		F		G		H	
Remaining endurance	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes	%MHT	Metric minutes
Subject No. 1	97	14.3	93	13.8	93	13.7	86	12.8	89	13.2	79	11.7	85	11.6	72	8.6
Subject No. 2	96	14.9	93	14.4	92	14.3	86	13.3	89	13.8	79	12.2	86	13.2	72	11.1
Subject No. 3	96	12.3	93	11.9	92	11.8	86	11.0	88	11.3	79	10.1	85	10.9	72	9.2
Subject No. 4	96	12.8	93	12.4	92	12.3	86	11.5	89	11.8	79	10.5	85	9.6	72	9.6
Subject No. 5	96	9.0	93	8.7	93	8.7	86	8.1	88	8.3	79	7.4	85	6.7	72	6.3
Subject No. 6	96	6.4	93	6.2	93	6.2	---	---	88	5.9	---	---	---	---	---	---
Subject No. 7	96	7.4	94	7.2	92	7.1	86	6.6	88	6.8	79	6.1	85	6.3	72	5.3
Subject No. 8	96	13.3	93	12.8	92	12.7	86	11.8	88	12.2	79	10.9	85	11.5	72	6.3
Mean ± s.d.	96.1 ±0.35	11.30 ±3.245	93.1 ±0.35	10.92 ±3.12	92.4 ±0.52	10.85 ±3.088	86.0 ±0.00	10.73 ±2.472	88.4 ±0.52	10.41 ±3.000	79.0 ±0.00	9.84 ±2.258	85.1 ±0.38	9.97 ±2.601	72.0 ±0.00	8.06 ±2.122

4.4.2 Discomfort ratings

As previously mentioned, the collection of information about the discomfort created by the holding of the experimental posture was aimed at identifying the parts of the body where the discomfort first reached a level so high as to force the subject to stop the effort.

On every occasion they held the posture to that limit (including the initial measurement of MHT), the subject identified the body sites where they experienced the worst discomfort. Table 4.7 presents a summary of that information, and each site has been related to the superficial muscle most likely to be affected by the unbearable discomfort. When preparing table 4.7, every mention to a particular site was accounted for , even though the subject might have referred to a region comprising more than one site.

Table 4.7 Body sites affected by the worst discomfort following the holding of both arms abducted to 60°, to the limit of endurance. Data from 68 trials.

Body site with the worst discomfort	Number of reports		Superficial muscle most likely to be affected
	Right arm	Left arm	
External aspect of arm, upper third	50	38	Medial deltoid
Posterior aspect of arm, upper third	14	9	Posterior deltoid
Posterior aspect of area between neck and acromion	20	12	Trapezius, descending part

4.5 Discussion

4.5.1 Degree of agreement between predicted recovery and observed recovery

A comparison of the information presented in tables 4.5 and 4.6 shows that in most of the cases Milner's model predicted a degree of recovery larger than that observed. Only in two cases (combination F for subject 8, combination H for subject 7) did the two values coincide.

It is convenient to indicate at this point that, except where otherwise indicated, all the statistical treatment of data was performed using the MINITAB software package (MINITAB, 1991) and the level of significance for all the tests carried out was set at $p= 0.05$. This statement also applies to the statistical procedures described through the remainder of the thesis.

To assess whether the difference between the predicted and the observed recovery was statistically significant, a t-test was carried out to compare the mean value of the 60 predicted recoveries (10.3 minutes) with the mean of the 60 observed recoveries (8.73). The test found that the difference between those mean values was significant: $t = 3.38$, on 113 d.f., with $p<0.001$. This result showed that Milner's model did in fact predict a degree of recovery significantly larger than what could be achieved by the subjects.

Another way of testing the concordance (or otherwise) of the predicted and the observed degree of recovery is by submitting their values to a linear regression analysis. In the ideal case that each observed value coincided with the corresponding prediction by the model, the best-fit regression line would have a slope coefficient of 1, and the coefficient of determination for the regression model would be a perfect 100%. When this procedure was applied to the observed and predicted values of percentage recovery (shown in table 4.5 and 4.6, respectively) obtained from the 8 experimental conditions investigated in the present study, the equation for the regression line was Predicted = 1.16 (Observed), with R^2=96.5%. Remarkably, the regression equation reported by Milner (1985) for the same procedure applied to the data he obtained from the posture he studied (a stoop with arms extended to the front) had the form Predicted = 1.01 (Observed), with R^2 equal to 57.6%. So, whilst the values predicted by the model for that posture were only an average of 1% larger than the observed ones, the fit for the regression line was only moderately good, which Milner attributed to the large variability of the data. Nonetheless, the coefficient 1.16 for the slope of the relationship between the observed and predicted values obtained in the present study, and the fact that the regression explained nearly all the variation present in the data (R^2= 96.5%) show that Milner's model does in fact overestimate (by an average of 16%) the recovery to be expected from the subjects who hold the posture with arms abducted at 60 degrees.

Since the model was actually tested in eight different conditions, it was necessary to find out whether the extent of the disagreement between the predicted and the observed recovery in each of those conditions was the same. To do this, the value of the observed recovery was subtracted from the predicted value and one-way analysis of variance was carried out on the difference (predicted - observed), with the combination as test factor. The recovery values used in this test were those expressed as percentage of the corresponding maximum holding time. The results of the procedure are shown in table 4.8.

Table 4.8 Analysis of variance on the difference between predicted and observed percentage recovery.

SOURCE	DF	SS	MS	F	p
COMBINATION	7	1553	222	1.13	0.356
ERROR	52	10164	195		
TOTAL	59	11717			

COMBINATION	N	MEAN	STDEV
A (25/25)	8	6.12	11.84
B (25/12.5)	8	14.87	20.02
C (50/50)	8	13.25	12.94
D (50/25)	7	11.00	14,50
E (75/75)	8	18.88	12.18
F (75/37.5)	7	6.57	13.87
G (100/100)	7	16.71	9.89
H (100/50)	7	3.86	13.92

According to this result, the variation in the average difference between predicted and observed recovery was not significantly different when compared between the eight experimental conditions ($F_{7,52} = 1.13$, $p = 0.356$). However, the small value of F is due mainly to the large error term, which reflects the fact that within each condition there was wide inter-subject variation in the difference between predicted and observed recovery. In consequence, obvious differences between mean values for experimental conditions, such as E and G for example, became masked by the variability between subjects.

As figure 4.1 shows, the considerable difference between the average degree of recovery predicted by Milner's model and the average of the recovery measured in each of the eight experimental conditions was well evident when the values presented in table 4.8 were plotted on a co-ordinate plane, with the length of the rest shown on the horizontal axis and the degree of recovery on the vertical axis (both expressed as % MHT). For the eight combinations of work and rest, the average value of the observed recovery (the points in the graph, illustrated with the corresponding s.d. bar), lay below the curve for the corresponding length of initial holding. This spatial relationship confirms that, on average, the model consistently predicted a degree of recovery larger than what the subjects could achieve. The four curves that appear on the graphic were drawn by calculating the degree of recovery that Milner's model would predict if the subjects were asked to perform initial holdings of length equal to 25%, 50%, 75% and 100% of their MHT, combined with periods of rest of length equal to 10%, 20%...90%, 100% MHT.

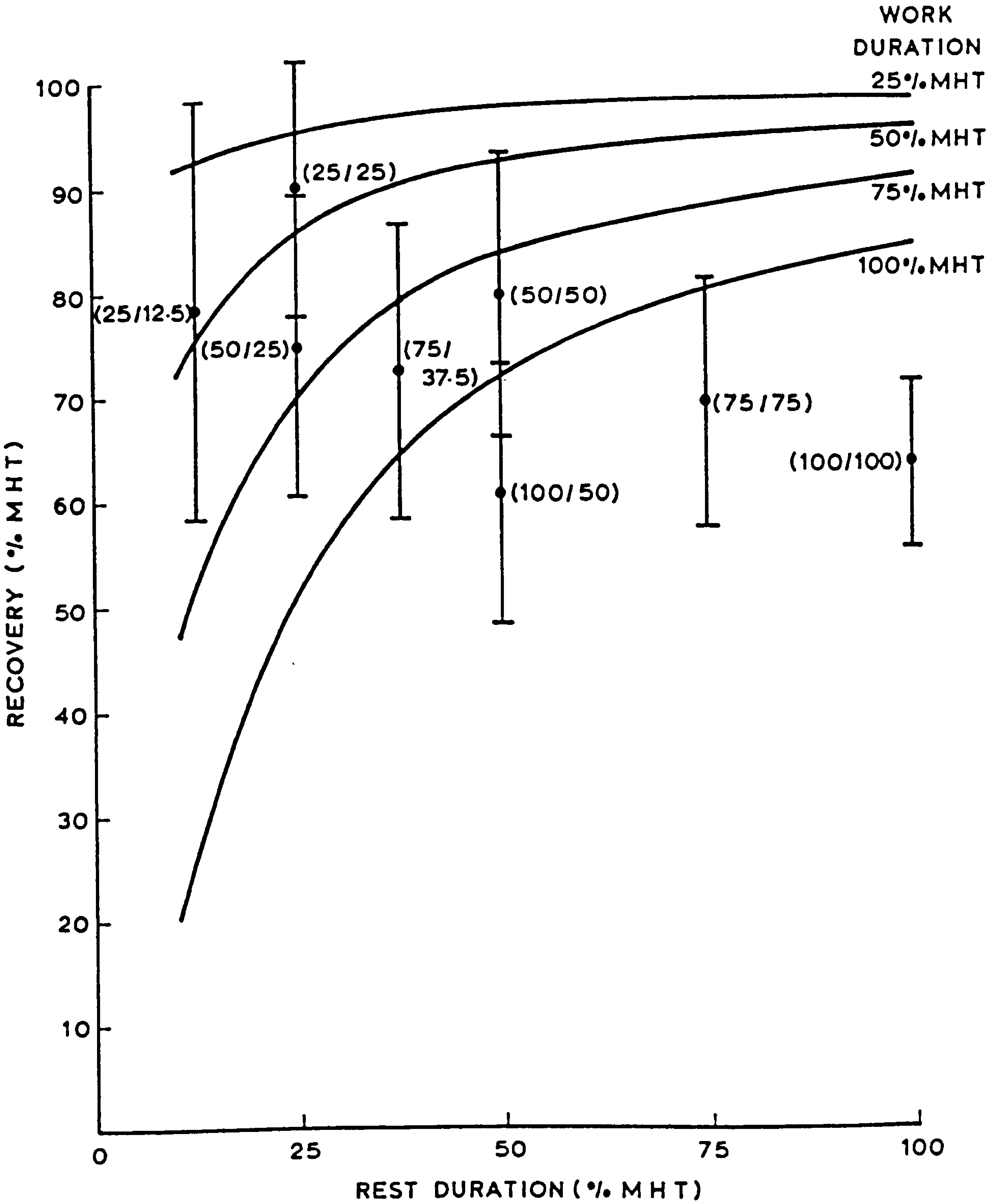


Figure 4.1 Plot showing the average degree of recovery measured in each of the eight experimental conditions (points and bars, for mean ± s.d.) and the average predicted by Milner's model (continuous curves). See text for details on the tracing of the curves.

Thus, the graphical evidence backs the result of the statistical tests practised on the data obtained in the present study, showing that the disagreement between the recovery predicted by Milner's model and the observed recovery was not attributable to chance. Therefore, all the evidence shows that the model is not valid for the posture with arms abducted at 60° that was used in the first experimental stage of the present investigation.

4.5.2 Other significant discrepancies between the present and Milner's studies

The mean value of the difference between predicted and observed percentage recovery presented in table 4.8 disagree with another finding made by Milner when he developed the model and tested its repeatability.

Milner reported that for every pair of conditions he studied where the same length of initial holding was combined with two different lengths of rest (e.g. 50% MHT/50% MHT and 50% MHT/25% MHT, or C and D in table 4.2), he found that the condition with the larger rest always resulted in a recovery that was closer to that predicted by the model than the other one. However, in the present study that was the case only for combinations A and B, with an initial holding's length equal to 25% MHT; for the other three pairs of combinations the relationship was the opposite, with the shorter rest resulting in recoveries that were closer to the predicted value. This fact is also evident in figure 4.1: whilst the point A (25/25) lay closer to the curve of 25%

MHT work duration than did the point B (25/12.5), for the other three pairs of experimental conditions (C-D, E-F and G-H) the relationship was reversed.

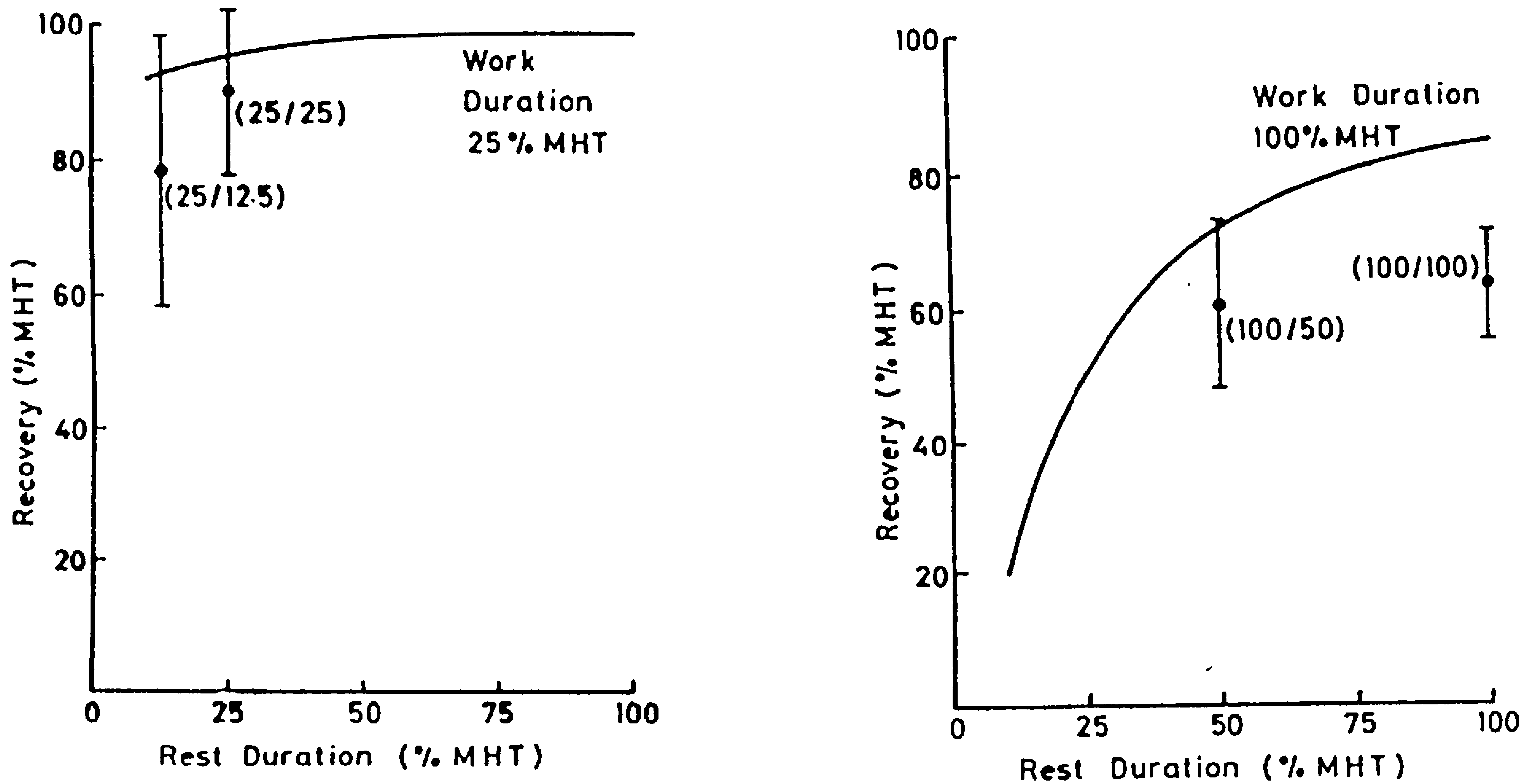
To make it easier to appreciate the difference in the response to conditions A and B as compared to conditions G and H, the sections of the plot in figure 4.1 corresponding to those two pairs of combinations were drawn on their own, and are shown in figure 4.2. Indeed, the 'zooming' onto the curve and points for 100% MHT revealed that the obvious difference in the length of rest for combinations G and H (G being double than H) was not reflected in the corresponding recovery which, on average, was practically the same for both conditions: 68.1 %MHT for G and 68.4 %MHT for H.

The most likely reason behind the fact that not always the larger rest led to the higher recovery was the variability within the subjects. The values presented in table 4.5 show that every subject had at least one instance where, for a given pair of combinations initial holding/rest, their recovery after the holding with the shorter rest was much larger than expected. Furthermore, it was the case that sometimes the subject 'over-recovered', that is, the observed recovery was larger than predicted The extent of the within-subject variability may be appreciated more clearly in table 4.9, which presents the size of the difference [predicted - observed (percentage) recovery] transformed into a percentage of the first one.

Table 4.9 Difference between predicted and observed recovery, expressed as a percentage of the first one. A minus sign indicates that the subject recovered beyond the prediction.

	Combination holding/rest							
Subject	A	B	C	D	E	F	G	H
1	3	23	44	40	27	28	36	-13
2	7	37	12	11	33	16	14	39
3	-2	-13	9	6	12	-22	27	-4
4	14	25	7	-1	44	24	3	27
5	3	10	-1	17	10	15	12	-6
6	-16	-18	6	--	5	--	--	--
7	14	19	15	-11	13	-3	16	0
8	27	41	23	26	30	0	31	-5

The values shown in table 4.5 for subject No. 3 are indeed a clear example of the within-subject variability of the degree of recovery. For every pair of combinations with the same length of initial holding, she did achieve higher recovery with the shorter rest, on three occasions recovering even beyond the predicted level. Subject No. 7 also behaved in very much the same way: only for the combinations with initial holding equal to 25% MHT she did recover better with the longer rest. Thus, the relationship between the degree of recovery reached by the subject and the extent of rest they were allocated when performing holdings of equal length is another aspect in which the present study differs from that by Milner (1985)



(A)

(B)

Figure 4.2 Plot of the average degree of recovery predicted by Milner's model (continuous curves) and the mean and s.d. (points and bars) of the actual value observed when the model was tested on 7 female subjects. The test conditions included an initial holding of length equal to 25% MHT (section A) and 100 %MHT (section B).

Another quite significant discrepancy between the two studies emerged when considering the approach taken by Milner in testing the applicability of the model to a posture different from that he used in its development. The test posture required the subjects to stand upright and locate their hands at a height equal to their own stature, and at a distance equal to arm reach. Six subjects had their maximum endurance to that posture measured, and then performed the combinations of initial holding lasting 33% and 66% MHT with a rest of length equal to 25%, 50% and 100% of the holding just performed (that is, between 8% and 66% MHT).

Contrary to what happened in the stooping posture with arms extended to the front that was used to develop the model, for the upright posture with the arms raised to place the hands at head level, the model tended to predict recoveries larger than those observed (the situation found in the present study). However, Milner applied a t-test to the difference between the predicted and the observed values, and it showed that the difference was non-significant: $t=0.969$, $p= 0.3375$ (Milner et al, 1986). Again, this result contrasts with the findings from the present study, for as has been shown already in the previous section, a t-test found a significant difference between the average observed recovery and the average predicted recovery ($t = 3.38$, on 113 d.f., with $p<0.001$).

4.5.3 Assessment of the exponential term in Milner's model

Thus, as has been shown in the two previous sections of this chapter (both statistically and graphically), the model to predict recovery from pure postural exertion proposed by Milner (1985) does not apply to the data obtained in the present study.

A review of the conceptual and experimental frame in which the testing of the model was conducted suggests that the weightier reasons for such result might lie with the differences between the postures investigated and the way they act on the anatomical structures involved. Clearly, the postures were designed to place the main stress on quite different body regions (lower back in Milner's study, shoulders in the present one), and whilst it might be true that the physiological phenomena that lead to fatigue occur in the same basic way, regardless of the body region being subjected to the effort, it is still possible that the anatomical relationships between skeletal and muscular structures in those most stressed parts of the body could in the end make a significant difference to the actual course of the fatigue. Another factor whose relevance must also be considered is that of the subject's gender, for again although one might not expect to find that male and female subjects react in significantly different manner to muscular loading, that has been found to be so (e.g. Takala et al 1993).

However, before assuming that the explanation for the non-adjustment between the results of this study and Milner's model is to be found only in those quite substantial factors just mentioned, it is worth considering whether it might lie with the model itself, and specifically with the structure of the equation which expresses it. This then gives rise to the question of whether all it is needed to improve the fit between the model and the empirical data is a slight modification to the form of that equation.

To assess that possibility, the data were submitted to a non-linear regression procedure, using the programme BMDPAR (Derivative-free nonlinear regression, BMDP 1988). This programme runs the data iteratively through a function of form specified by the user, modifying the parameters that he indicates. The user may either limit the scope for the modification of the function implemented by the programme, by setting a cut-off point in the values of the parameters that are being tested, or allow the programme to perform as many iterations as required until the fit between data and function cannot be improved any further. Either way, the programme communicates the result of the procedure by producing a report showing how many iterations were carried out, the value(s) of the parameter(s) under test which produced the best fit between the non-linear model and the data, values of the mean and variance (with d.f.) for the variable being calculated, the estimated mean square error (with the corresponding degrees of freedom), and the value of the parameter 'pseudo r^2' which expresses the goodness of fit between the data fed to the programme and the model under evaluation. The value of the parameter pseudo r^2 is in the last instance the most important piece of the information shown in the report, for when the programme founds that the non-linear model being assessed fits the data less well than the mean, then that parameter will take a negative value.

Let's remind ourselves of the equation that expresses Milner's model:

$$\text{Recovery} = (\text{MHT} - \text{HT}) + \text{HT}[e^{-0.164(\text{HT}/\text{Rest})}]$$

The only parameter in this equation that the BMDPAR programme could modify from one run to the next was the value of the constant (-0.164) in the exponential term. Assuming that a truly slight modification to this parameter would be enough to improve significantly the fit between the model and the values of recovery observed during the trials (expressed as %MHT), the programme was constrained to assess values in the interval between -0.163 and -0.165. The programme found that the best fit was for the value -0.163. However, the value of pseudo r^2 associated to that result was -96.041, which showed that the fit between the new equation, with the constant in the exponential term changed from -0.164 to -0.163, was still well away from being the optimum.

This led to a second attempt, this time starting with the constant set at -0.164 and allowing the programme to make modifications until no further improvement could be achieved. The best possible fit was found after ten iterations, the constant having reached a value of -1.108. The value of pseudo r^2 had now changed to -48.397, indicating that the fit between the model and the experimental data was still far worse than the fit between the model and the mean value of the data. These result showed then that, in order to fit to the experimental data obtained during the present study, the model required a deeper modification than just changing the power of the exponential term.

This last finding is indeed quite significant, since Milner (1985) asserted (page 160) that his was "a model in a form which agrees with medical and physiological research". With this, Milner suggested that the exponential term in the equation incorporated the influence of the physiological factors present in the exertion of pure postural nature. However, it is indeed difficult to imagine that the physiology of the fatigue for the posture used in this study could be so different from that involved in the posture Milner used in his study. This consideration leads straight into the question of whether it is the model in itself where the problem lies. To answer this question (albeit at the most elementary level) it is necessary to review the assumptions on which Milner's model was based. An obvious starting point for that review is the issue of whether the maximum holding time (which Milner used as the cornerstone for the model) is in fact as consistent as Milner thought it to be. No further elaboration on this point will be made here, since the issue will be dealt with elsewhere in this thesis.

4.5.4 Further information provided by the discomfort ratings

Before bringing this chapter to an end, it is worth considering some interesting aspects of the information provided by the subjects in relation to the discomfort provoked by the postural exertion.

The first of those aspects refers to the level of discomfort experienced by the subjects at the end of the initial holding, and whether some of it remained following the rest pause. The relevant information is shown in table 4.10. In general, the values followed the trends that could be expected, since the longer the initial holding, the higher the discomfort reported by the subject at the end of it. Also, for the pairs of combination with the same length of initial holding, in most of the cases the shorter pause for rest led to the subject returning a higher rating of discomfort at the start of the second holding.

The information given in table 4.10 also bears out another manifestation of the variability between subjects: they returned very different discomfort ratings at the end of periods of exertion that, theoretically at least, represented the same relative demand on their endurance capacity. For example, at the end of the initial holding in combination A, whilst subject No. 8 sensed the discomfort as 'just noticeable', a rating of 0.5, there were four other subjects who perceived it as 'moderate' already, a rating of 3. In contrast, there was no noticeable variability within-subject: they all appeared to handle the rating scale in a consistent manner, returning fairly similar values at the end of exertion periods which were of the same length.

Table 4.10 Level of discomfort and sites where it appeared at the end of the initial holding and at the beginning of the second one.

Combination	A (25/25)		B (25/12.5)		C (50/50)		D (50/25)		E (75/75)		F (75/37.5)		G (100/100)		H (100/50)	
Discomfort rating	End of hold I	Start of hold II	End of hold I	Start of hold II	End of hold I	Start of hold II	End of hold I	Start of hold II	End of hold I	Start of hold II	End of hold I	Start of hold II	End of hold I	Start of hold II	End of hold I	Start of hold II
Subject No. 1	3F	0	2F	0.5F	6F	0.5F	6F	2F	8F	0	9F	1E,F	10F	1E,F	10F	0
Subject No. 2	2F	0	1E,F	0.5F	5F	0.5E	4F	1F	7F	0	8F	0	10F	0.5E,F	10F	4E
Subject No. 3	3E	0	2E	0	5E	0	5E	0.5E	10F	0	8E	0	10E	0.5F	10E,F	0
Subject No. 4	3F	0.5F	3F	2F	7F	1F	6F	2F	10F	1F	9F	2F	10F	3F	10F	2F
Subject No. 5	2F	0	2F	0	4F	0.5E	1F	0.5E,F	9F	0	8F	1F	10E,F	0	10F	0.5E,F
Subject No. 6	1E	0	3E,F	0	4E,F	0	---	---	9,EF	0	---	---	---	---	---	---
Subject No. 7	3F	0.5F	5F	1F	9F	1F	8F	2F	9F	1F	10F	1F	10F	2F	10F	1F
Subject No. 8	0.5E,F	0.5E	1F	0	4F	0.5E	5F	0	8F	0	7F	0	10F	0	10F	0

NOTE: The letters next to the value of the discomfort rating identify the body regions (shown in figure 3.8) where it was being experienced by the subject.

As already shown in table 4.7, the muscles that appeared to bear the brunt of the loads created by the posture used in this study were the descending part of the trapezius and the medial and the posterior parts of the deltoid. However, the comparison of the information contained in tables 4.7 and 4.10 highlighted an interesting fact: at the end of the initial holding all the subjects reported the higher discomfort in the areas corresponding to the medial and posterior parts of the deltoid muscle, no mention was made to the region covered by the trapezius muscle (table 4.10). Nevertheless, by the end of the whole sequence of two holdings, the trapezius muscle appeared to be another site of extreme discomfort, which was evidenced by the number of mentions it received (table 4.7). However, further comment on this matter will be withheld until chapter 6, where the results of the collection of discomfort ratings during the second series of experiments will be presented.

Finally, besides permitting the identification of the body regions where the experimental posture imposed the larger loads, the discomfort ratings can also be used to investigate the onset of fatigue as the holding time progresses. This, however, was not the main purpose of the trials being reported in this chapter. The second series of experiments in the present investigation was designed to look at the issue at length, and the results of that search will be presented on their own in chapter 6.

4.6 Conclusions

The first stage of the experimental work in this study was undertaken with the aim of fulfilling the first of the specific objectives expressed in chapter 3: finding out whether the equation developed by Milner (1985) to predict the level of recovery after holding to exhaustion a standing stooped position, could be used to predict the recovery from fatigue provoked by holding to exhaustion an upright standing posture that loaded the shoulders. The evidence presented in this chapter shows that, definitely, such was not the case, Milner's model does not translate from one posture to the other.

In consequence, the failure of Milner's model had a significant impact on the expressed intentions of this investigation, since rather than attempting the extension of the model to a wider set of experimental conditions, the aim now shifted to testing the safety of the key assumption on which the model was based. Such is the matter for the next chapter.

CHAPTER 5

THE MAXIMUM HOLDING TIME AS A MEASURE OF ENDURANCE TO POSTURAL LOADING

5.1 Introduction

It has been shown in chapter 4 that the model developed by Milner (1985) to predict the recovery from the exhaustion experienced at the end of a single sequence of two holdings of a posture cannot be transferred between postures. This finding, which goes against what Milner affirmed, raised several quite substantial questions. Although these have been expressed already in the last chapter, it is certainly worth repeating them here briefly, ranked according to the increased difficulty posed by the search for the corresponding answers:

a) Is the failure due to the way in which the model was mathematically structured?;

b) Is it perhaps that Milner used a basic building block (the maximum holding time of postural loading) which is inherently flawed?;

c) Is the failure of the model simply a reflection of the intrinsic differences between the postures;

d) Or is it rather the result of differences in the way that male and female subjects react to postural loading?

In fact, question a) was addressed in the discussion to chapter 4, leading to the conclusion that even though Milner asserted that the structure of the model tallies with most of the research findings about the physiology of fatigue in isometric exertion, this may well not be the case. However, an in-depth search for an answer to this issue is most definitely beyond the boundaries of the present investigation and, for the time being, no further attempt may be made to find it.

However, before thinking of moving into the search for answers to questions c) and d), it is necessary to address the issue raised by question b). Quite simply, this issue revolves around the consideration of whether Milner's assumption regarding the maximum holding time (MHT) as a sufficient basis for the model was safe. Milner himself acknowledged (Milner 1985, p 223) that his subjects differed largely in respect of their MHT, a circumstance that in turn affected the quality of the predictions yielded by the model. Nevertheless, in the course of his work he did not endeavour specifically to try and find out how consistent the maximum holding time was. This is in fact tantamount to asking whether a person will reach the same maximum holding time when this is measured on different occasions. Besides, since he limited his study to a single posture and to male subjects only, neither could Milner consider the influence that the gender of the subject, or the variations in the posture, could have on the maximum holding time.

This chapter reports on the work carried out to test the nature and extent of those effects. The experimental design was one of repeated measurements on both male and female subjects who held to the limit a standing posture whilst their arms were placed at three abduction angles.

In addition to testing for the effects of the gender of the subject and the change in the abduction angle, attempts were made to find a regression model that would explain the variations of the maximum holding time as a function of those variables, of a number of the anthropometric characteristics of the subjects and of the biomechanical moments imposed on the shoulder joint by the three postures studied.

Those attempts were part of the main experiment, which was also designed to study the change in the subjective perception of discomfort and to monitor changes in the myoelectrical signal of three superficial muscles known to be heavily engaged during the postural exertion involved. The procedures applied in those studies and the corresponding results will be the matter of chapters 6 and 7, respectively.

5.2 Procedures

5.2.1 Subjects

The sample consisted of ten subjects, five male and five female, whose details are given in table 5.1. This size of sample was decided on the basis of the values of MHT collected during the earlier trials with sequential holdings at 60°; it was designed to allow the calculation of the MHT within a 95% confidence interval, with a precision of ± 5%.

Table 5.1 Details of the subjects who took part in the main experiment.

Subject's identifier	Gender	Age (years)	Weight (kilograms)	Stature (mm)	Shoulder height (mm)	Arm length (mm)	Forearm length (mm)
No. 1	Female	19	65.8	1570	1272	629	420
No. 2	Female	19	77.2	1695	1403	719	490
No. 3	Female	22	65.3	1735	1433	709	484
No. 4	Female	21	62.4	1693	1366	733	480
No. 5	Female	24	53.6	1661	1352	735	490
No. 6	Male	21	59.6	1791	1464	767	534
No. 7	Male	18	76.4	1795	1474	750	520
No. 8	Male	21	57.5	1686	1383	725	492
No. 9	Male	23	69.5	1790	1464	752	520
No. 10	Male	22	60.2	1711	1380	715	480
	Mean ± s.d.	21.0 ± 1.88	64.75 ± 7.777	1712.7 ± 69.78	1399.1 ± 62.62	723.4 ± 37.81	491.0 ± 31.39

The criteria for selection of the ten subjects, and the pre-recruitment process of screening and information given to them have been described already in chapter 3.

5.2.2 Methods

The experimental postures used in this study involved abduction angles of 30°, 60° and 90°. This choice of postures had the aim of covering the most frequently occurring postures within the normal range of movement in active arm abduction, which has been reported to be 90° (Lucas 1973). In addition, those postures are commonly found in the everyday work of large numbers of people employed in industries with a heavy leaning towards assembly tasks, such as shoemaking (Serratos-Pérez and Mendiola-Anda, 1993).

The experimental design for the trials consisted of the measurement, on three separate occasions, of the maximum holding time for each of those three postures. The number of measurements was limited to three mainly for reasons of availability of time and resources; however, three is also the minimum number of replicates necessary to identify a trend in the behaviour of a variable.

The subjects completed the nine trials (3 postures x 3 replicates) in random order. However, they attended the laboratory on a total of ten occasions. During the first one they were weighed and measured up and, subsequently, the settings of the experimental platform were adjusted so as to get each subject in the correct experimental postures. These measurements and adjustments were performed following the procedures already described in chapter 3.

The other nine sessions were devoted to the experimental work proper. During these, whilst the subject held the corresponding posture, at least once every minute the researcher checked that they had not deviated significantly from it. If that was the case, the subject was asked to make the necessary adjustments, so that they returned to the intended posture

5.3 Results and statistical analysis

5.3.1 Length of the maximum holding time

Table 5.2 shows the maximum holding times achieved by each subject during each of their nine trials. At 30°, the values ranged between 451 seconds (3rd trial by subject No. 1) to 3623 seconds (3rd trial by subject No. 9); at 60° the range was between 330 seconds (1st trial by subject No. 1) and 2204 seconds (3rd trial by subject No. 9) and at 90°, the range extended from 175 seconds (2nd trial by subject No. 1) to 988 seconds (2nd trial by subject No. 9).

In contrast with the results reported in chapter 4, the holding times achieved during the main experiment will not be compared with those reported by Milner (1985), who measured MHT in metric minutes. Therefore, the times measured during the main experiment will be expressed in seconds.

Table 5.2 Maximum holding time (seconds) to postural loading of the shoulders by abduction of the upper arms in the coronal plane.

Subject	Trials with abduction at 30°			Trials with abduction at 60°			Trials with abduction at 90°		
Identifier	1st	2nd	3rd	1st	2nd	3rd	1st	2nd	3rd
No. 1	467	519	451	330	347	389	180	175	206
No. 2	1357	1409	1469	519	568	638	330	419	365
No. 3	1555	2091	1845	702	959	998	452	487	515
No. 4	659	578	641	460	405	407	227	337	392
No. 5	879	862	1157	481	505	623	349	383	383
No. 6	2729	2820	2900	1555	1250	1826	446	531	524
No. 7	875	746	1042	581	691	679	325	387	379
No. 8	537	887	1074	473	632	572	343	285	390
No. 9	2729	2673	3623	1473	1891	2204	933	988	873
No. 10	2978	3607	3577	1917	1903	1948	752	577	515
Mean ± s.d.	1476.5 ±983.27	1619.2 ±1102.37	1777.9 ±1177.38	849.1 ±570.40	915.1 ±581.60	1028.4 ±691.77	380.7 ±156.24	456.9 ±220.62	454.2 ±175.10

5.3.2 Effects on the MHT of the variation in the posture and the gender of the subject

The data presented in table 5.2 show that the increase of the abduction angle had the obvious effect of shortening the maximum holding times. For the whole sample of ten subjects, the mean value and standard deviation at each angle were: at 30°, 1624 seconds (s.d. = 1060); at 60°, 932 seconds (s.d. = 602); at 90°, 448 seconds (s.d. = 204). The analysis of variance showed that the difference between the mean values was highly significant, F-value was 20.64, with $p<0.001$.

A visual inspection of the data presented in table 5.2 makes evident that the maximum holding time has a large variability between subjects. However, the data also show that the increase of the abduction angle had the effect of reducing the extent of that variability. The coefficients of variation were 0.653 for the measurements performed at 30°, 0.646 for those at 60° and 0.455 for those at 90°. Judging by these values, the increase in the abduction angle provoked a reduction in the dispersion of the data, and this effect was more evident in the increase from 60° to 90° than in that from 30° to 60°. The size of the coefficients of variation confirms the significance of the degree of variability between the subjects.

To assess the influence of the gender of the subject on the length of the maximum holding time, the data presented in table 5.2 were grouped according to gender, and the respective values of mean, standard deviation and coefficient of variation were calculated. Table 5.3 shows the results of these calculations.

Table 5.3 Summary statistics for the maximum holding time (measured in seconds) achieved by male and female subjects during trials of arm abduction.

Subject gender	Trials at 30°			Trials at 60°			Trials at 90°		
	mean	s.d.	c.v.	mean	s.d.	c.v.	mean	s.d.	c.v.
Female	1063	532	0.500	555	203	0.366	347	107	0.308
Male	2190	1171	0.535	1306	635	0.486	550	230	0.418
z-value	5.55 ($p<0.001$)			7.56 ($p<0.001$)			5.37 ($p<0.001$)		

Table 5.3 also includes the results of the z-tests carried out to compare the average endurance exhibited by the two groups of subjects. These tests showed that, at the three abduction angles, the endurance of the male subjects was significantly larger than that of the females. Interestingly, the extent of the reduction in the endurance capacity that accompanied the increase of the abduction angle was approximately the same for male and female subjects. Thus, the average endurance of the female subjects at 60° was 50% of what they achieved at 90°, and for the male subjects it was 60%. At 90° the female subjects endured on average 33% of what they did at 30°, whilst the males endured 25%.

However, the coefficients of variation included in table 5.3 show that only when the arms were held at the lowest abduction angle was the variability between subjects roughly of the same size for both male and female subjects, the corresponding coefficients of variation being 0.535 for males and 0.500 for females. At the other two abduction angles the maximum holding time varied more widely for the male subjects than it did for the females, the coefficients of variation at 60° were 0.486 (males) and 0.366 (females); at 90° they were 0.418 (males) and 0.308 (females). These values also show that the reduction of variability that accompanied the increase of the abduction angle was more evident for the female subjects than it was for the males.

5.3.3 Within-subject variability of the Maximum Holding Time

In order to assess the variability exhibited by the endurance of each subject to the postural loads, the mean, standard deviation and coefficient of variation of the maximum holding times achieved during the three trials at each abduction angle were calculated. The results are presented in table 5.4. It is worth noting that the wide variations between subjects were as evident in the averaged values of MHT as they were in the individual values, which have been presented already in table 5.2.

Table 5.4 Summary statistics of the maximum holding time (seconds) achieved by each subject during the three trials performed at each abduction angle.

	Trials at 30°			Trials at 60°			Trials at 90°		
Subject	Mean	Standard deviation	Coefficient of variation	Mean	Standard deviation	Coefficient of variation	Mean	Standard deviation	Coefficient of variation
No. 1	479	36	0.075	355	30	0.084	187	17	0.090
No. 2	1412	56	0.040	575	60	0.104	371	45	0.121
No. 3	1830	268	0.146	886	161	0.182	485	32	0.066
No. 4	626	42	0.067	424	31	0.073	319	84	0.263
No. 5	966	166	0.172	536	76	0.142	372	20	0.054
Female sub-sample	1063	532	0.500	555	203	0.366	347	107	0.308
No. 6	2816	86	0.031	1544	288	0.187	500	47	0.094
No. 7	888	148	0.167	650	60	0.092	339	53	0.156
No. 8	833	273	0.328	559	80	0.143	339	53	0.156
No. 9	3031	516	0.170	1856	367	0.198	931	57	0.061
No. 10	3387	355	0.105	1923	23	0.012	615	123	0.200
Male sub-sample	2190	1171	0.535	1306	635	0.486	550	230	0.418
Overall sample	1624	1060	0.653	932	602	0.646	448	204	0.455

The coefficients of variation included in table 5.4 show that the within-subject variability was also important. However, only for subject No. 1 was the size of the variation roughly the same at the three experimental conditions. For the rest of the subjects the variability of their maximum holding times changed widely between conditions, but neither was this change characterised by a consistent trend. For example, subject No. 4 had a rather low coefficient of variation which was similar at 30° and 60° and then it went up by nearly fourfold for the measures at 90°. However, the opposite happened for subject No. 5, whose data had coefficients of variation at 30° and 60° that were between three and four times as large as the one at 90°.

Nevertheless, despite the important variations between and within-subject, one-way analysis of variance on the whole sample found that the average of the maximum holding times achieved during the three trials was not significantly affected by the order in which those trials were completed. The results of the analysis of variance are presented in table 5.5. The values of p included in the table show that the differences in MHT traceable to the order of the trials were far from significant at the three abduction angles. Indeed, although the average MHT for the whole sample increased from trial to trial (except for the holdings at 90°), this was not so at individual level, since the data presented in table 5.2 show that not a single individual did in fact increase their endurance from trial to trial and for all three angles

Table 5.5 Results of one-way ANOVA to assess the order effect on the MHT (seconds) of the ten subjects during their three trials at each abduction angle.

	Abduction angle					
	30°		60°		90°	
Trial order	Mean endurance	Standard deviation	Mean endurance	Standard deviation	Mean endurance	Standard deviation
First	1483	993	849	570	434	235
Second	1619	1102	915	582	460	221
Third	1778	1177	1028	692	454	175
F-value	0.18		0.22		0.04	
p-value	0.835		0.807		0.965	

On the other hand, the consistency of the maximum holding time as a measure of the endurance to postural loading was also evident in the results obtained during the first experimental stage of the present investigation, when to test Milner's model the 8 female subjects who took part in the trials were asked to hold the posture with arms abducted at 60°. They had to perform three maximum holdings, one at the beginning of the experiment to set the duration of subsequent exertions and two more as part of the testing procedure, which were completed in random order. Table 5.6 shows the MHT achieved by each subject on those three occasions. It is convenient to remember that these were measured in metric minutes, to allow their comparison with Milner's results.

Table 5.6 Length of three maximum holdings of a standing posture with the arms abducted at 60°, performed during the testing of Milner's model.

Subject	First maximum holding time (metric minutes)	Second maximum holding time (metric minutes)	Third maximum holding time (metric minutes)
No. 1	14.8	13.7	12.0
No. 2	15.5	15.4	15.4
No. 3	12.8	12.8	12.8
No. 4	13.3	11.3	13.3
No. 5	9.4	7.9	8.8
No. 6	6.7	---	---
No. 7	7.7	7.4	7.4
No. 8	13.8	13.5	8.7
Mean ± s.d.	11.75 ± 3.349	11.71 ± 3.034	11.20 ± 2.935

The information presented in this table shows that, although there were variations in the length of the MHT achieved by each subject during the three measurements, the average values were remarkably close to each other. This was clearly demonstrated by the result from one-way ANOVA performed on the data, which were $F_{2,19} = 0.07$, $p < 0.05$. Indeed, some of the individual results were quite interesting, since the subjects returned subsequent lengths of MHT which were either a perfect replicate of the length they achieved during the first measurement (subjects No. 2 and 3) or quite close to it (subject No. 7). This is even more remarkable considering that the subjects were not aware that they were attempting to reach the 100% of their original MHT.

5.3.4 Influence of the experimental design and the individual characteristics on the variability of the maximum holding time for postural loading

The variations in an individual's MHT that were observed in the present trials may be attributed to a variety of factors. Some of those factors pertained to the experimental design itself: the gender of the subject, the repeated trials, and the change in the abduction angle. Some other factors were inherent in the subjects themselves by virtue of their bodily dimensions: weight, height, shoulder height, arm length and forearm length.

The change in the abduction angle in turn meant that the abduction moments acting on the shoulder joint also changed. Those moments were calculated using the 3-D biomechanical analysis software developed by Tracy (1990). This programme calculates the moments acting on the major body joints during static work, once the user has fed-in information about the subject's body weight and stature, as well as the size and direction of the forces acting upon or being exerted through the joints. The abduction moments acting on each shoulder were calculated for each subject at the three abduction angles. A t-test on the difference [left -right] showed that there was no significant difference between the moment acting on the right arm and that acting on the left arm: $t= -0.21$, $p= 0.83$, d.f.= 177. The average abduction moments for the whole sample of ten subjects at each angle (calculated for the right arm) were 4.3 N-m at 30°, 5.8 N-m at 60°, and 6.3 N-m at 90°. One-way ANOVA on these values found the difference to be highly significant: $F_{2,87} = 68.30$, $p<0.0001$.

To determine the influence of the experimental and individual-related factors on the variation in the maximum holding time, a regression analysis was performed on the data collected during the 90 trials. The procedure was carried out using the programme for regression analysis of MINITAB Release 8 (MINITAB, Inc 1991).

As a first step into the regression analysis, the programme was asked to find what combination of the variables believed to have an influence on the variation in MHT would explain the largest proportion of that variation. The variables fed-in to the programme were (with their code name shown between brackets): the abduction angle (ANGLE), the weight of the subject (WEIGHT), their height (HEIGHT), shoulder height (SHOULHT), arm length (ARMLGT), forearm length (FARMLGT), the abduction moments acting on the right arm (ABDMOMRA) and on the left arm (ABDMOMLA), and the repetition of the trial (REPT). All the variables related with the length of body segments were expressed in metres; the subject's weight was in kg; the abduction moments in N-m; the angle was given values of 1 (for 30°), 2 (for 60°) and 3 (for 90°) and the repetition was also expressed as 1, 2 or 3.

The outcome of the programme informed about the regression models that, using an increasing number of variables (1, 2, 3...9) would explain the largest proportion of the variation in MHT. The second step in the analysis was then to find the regression equation for the model with the highest explanatory power. This two-step procedure was first applied to the data for the whole sample, and subsequently to the data for the subjects of each gender.

When the data for the whole sample were fed-in to the programme, it produced the following report:

Best Subsets Regression of MHT (for the overall sample)

```
                                                 A A
                                           S   F B B
                                       W H H A A D D
                                     A E E O R R M M
                                   R N I I U M M O O
                                   E G G G L L L M M
                                   P L H H H G G R L
                                   T E T T T T T A A
            Adj
Vars  R-sq  R-sq  C-p     s
1     31.8  31.1  50.4  711.98       X
2     48.5  47.3  19.1  622.63       X   X
3     52.9  51.3  12.2  598.54       X   X X
4     56.4  54.4   7.2  579.34       X X X   X
5     58.3  55.8   5.4  570.00       X X X X X
6     59.3  56.3   5.5  566.62       X X X X X X
7     59.9  56.5   6.2  565.69     X X X X X X X
8     60.0  56.0   8.1  568.71     X X X X X X X   X
9     60.0  55.5  10.0  571.94     X X X X X X X X X
```

The statistic C-p that appears in the report reflects both the precision and the goodness of fit of the regression model built on the combination of variables to which it refers. A small value of C-p indicates a model which is relatively precise (i.e. has small variance) whilst a value of C-p that is close to the number of parameters in the model means that such model fits the data well. According to the MINITAB reference manual, (page 7-20) the best subset of variables is the one which simultaneously has the larger adjusted R^2 (Adj R-sq in the report) and a value of C-p which is at the same time as small as possible and as close as possible to the number of parameters in the model. Choosing such subset of variables will lead to the regression model with the smallest mean square error, which in turn reflects in the lowest value of s, the estimate of the variance of the variable under study. In the case of the analysis for the whole sample, that means the regression model that incorporates the variables repetition, angle, height, weight, shoulder height, arm length and forearm length, leaving out the value of the abduction moment in both shoulders. This model explains 56.5% of the variation in the maximum holding time for postural loading.

The equation for such model, with the corresponding table of coefficients and analysis of variance was obtained using the regression facility of the same statistical package. The result was as follows:

The regression equation is (for the overall sample)

MHT = - 6005 + 82.4 (REPETITION) - 589 (ANGLE) - 23.9 (WEIGHT)
+ 24282 (HEIGHT) - 18255 (SHOULDER HEIGHT)
- 21798 (ARM LENGTH) + 18920 (FOREARM LENGTH)

Table of coefficients:

Predictor	Coef	Stdev	t-ratio	p
Constant	-6005	2624	-2.29	0.025
REPT	82.40	73.03	1.13	0.262
ANGLE	-589.27	73.03	-8.07	0.000

```
WEIGHT        -23.91     11.30    -2.11     0.037
HEIGHT         24282      5690     4.27     0.000
SHOULHT       -18255      7529    -2.42     0.018
ARMLGT        -21798      7918    -2.75     0.007
FARMLGT        18920     13340     1.42     0.160

s = 565.7     R-sq = 59.9%     R-sq(adj) = 56.5%

Analysis of Variance

SOURCE        DF        SS          MS          F        p
Regression     7   39202324    5600332    17.50    0.000
Error         82   26240322     320004
Total         89   65442648
```

An inspection of the p-values that appear in the table of coefficients shows that the variables with a significant relationship (i.e. p<0.05) with the maximum holding time were (listed in decreasing order of significance): the abduction angle, the subject's height, the arm length, the shoulder height and the subject's weight.

When the lengths of MHT achieved by the female subjects were analysed in search of the best subset of variables, the programme reported the existence of an extremely strong correlation between some of the variables that had been fed-in to the process, which made it impossible to complete the required analysis. To solve this problem, it was necessary to reverse the order of the stages and proceed first to find the regression equation, since this allows to identify which are the variables which are significantly correlated. The programme found that it was the length of the forearm the variable which impeded the process of finding the best subset of variables on which to build the regression model.

Removing the length of the forearm from the input to the programme allowed it to go ahead and find the best subsets as follows:

Best Subsets Regression of MHT (for the sub-sample of female subjects)

```
                                              A A
                                          S   B B
                                      W H H A D D
                                    A E E O R M M
                                  R N I I U M O O
                                  E G G G L L M M
                                  P L H H H G R L
                                  T E T T T T A A
            Adj
Vars  R-sq  R-sq  C-p     s
1     43.8  42.5  68.6  338.51      X
 2    70.7  69.3  18.2  247.34      X     X
3     74.2  72.3  13.4  235.11          X X   X
4     80.1  78.2   3.7  208.67      X   X X   X
5     81.0  78.6   4.0  206.52    X X   X X   X
6     81.5  78.6   5.1  206.69    X X   X X X X
7     81.5  78.1   7.0  209.15    X X X X X X X
8     81.6  77.5   9.0  212.02    X X X X X X X X
```

This result showed that for the trials performed by the female subjects the combination of variables with the largest explanatory power was formed by the abduction angle, the subject's height, the shoulders height, the length of the arm, the abduction moment acting on the right arm and the repetition of the holding being performed. These variables were then used to find the regression equation for the corresponding model, and this was the result:

The regression equation is (for the sub-sample of female subjects)

MHT = - 1966 + 51.1 (REPETITION) - 203 (ANGLE) - 88868 (HEIGHT)
+ 14355 (SHOULDER HEIGHT) - 1431 (ARM LENGTH)
- 155 (ABDUCTION MOMENT IN RIGHT ARM)

Table of coefficients:

Predictor	Coef	Stdev	t-ratio	p
Constant	-1966	1342	-1.46	0.151
REPT	51.07	37.74	1.35	0.184
ANGLE	-202.94	57.96	-3.50	0.001
HEIGHT	-8868	3628	-2.44	0.019
SHOULHT	14355	3316	4.33	0.000
ARMLGT	-1431	1478	-0.97	0.339
ABDMOMRA	-155.02	44.00	-3.52	0.001

s = 206.7   R-sq = 81.5%   R-sq(adj) = 78.6%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	6	7148210	1191368	27.89	0.000
Error	38	1623390	42721		
Total	44	8771600			

The table of coefficients provided with the regression equation shows that the variables significantly related to MHT were (in order of decreasing significance) the shoulder height, the abduction moment on the right shoulder, the abduction angle and the subject's height.

The proportion of the variance of MHT for the female subjects explained by the model with the best subset of variables (angle, subject's height, shoulder height, arm length, moment on the right arm and repetition) was 78.6%, which is 22% higher than the explanatory power of the model with the best subset for the whole sample.

When the results of the trials performed by the male subjects were submitted to the search for the best subset of variables, the programme again found that there was a strong correlation between variables which made it impossible to find the best combination for the building of the regression model. The regression facility of the programme traced the problem to two variables: the length of the forearm and the abduction moment on the left shoulder. Once those variables were excluded from the search process, this went ahead, producing the following result:

Best Subsets Regression of MHT (for the sub-sample of male subjects)

```
                                            A
                                        S   B
                                    W H H A D
                                  A E E O R M
                                R N I I U M O
                                E G G G L L M
                                P L H H H G R
                                T E T T T T A
            Adj
Vars  R-sq  R-sq  C-p    s
1     44.1  42.8  78.3  772.02    X
2     49.4  47.0  68.9  743.01        X     X
3     71.8  69.7  23.3  561.88    X   X X
4     80.0  78.0   7.7  478.72    X X   X X
5     81.8  79.5   5.8  462.37    X X X X X
6     82.7  79.9   6.0  457.39  X X X X X X
7     82.7  79.4   8.0  463.49  X X X X X X X
```

The regression equation that includes the variables which formed the best subset was calculated next, with the following result:

```
The regression equation is (for the sub-sample of male subjects)
MHT = 6718 + 114 (REPETITION) - 821 (ANGLE)- 1067 (WEIGHT)
         - 58632 (HEIGHT) + 490809 (SHOULDER HEIGHT)
         - 721787 (ARM LENGTH)

* NOTE *   WEIGHT is highly correlated with other predictor variables
* NOTE *   HEIGHT is highly correlated with other predictor variables
* NOTE *  SHOULHT is highly correlated with other predictor variables
* NOTE *   ARMLGT is highly correlated with other predictor variables

Table of coefficients
Predictor      Coef       Stdev      t-ratio        p
Constant       6718       5751        1.17        0.250
REPT          113.73     83.51        1.36        0.181
ANGLE        -820.57     83.51       -9.83        0.000
WEIGHT       -1067.3     255.8       -4.17        0.000
HEIGHT       -58632     29452        -1.99        0.054
SHOULHT      490809    135485         3.62        0.001
ARMLGT      -721787    178053        -4.05        0.000

s = 457.4      R-sq = 82.7%    R-sq(adj) = 79.9%

Analysis of Variance

SOURCE      DF       SS          MS          F        p
Regression   6   37879024    6313170     30.18    0.000
Error       38    7949657     209201
Total       44   45828680
```

Thus, as may be seen from the table of coefficients for the regression equation, the regression model with the largest explanatory power for the results from the trials performed by the male subjects incorporated the variables (listed in decreasing order of significance) abduction angle, subject's weight, arm length, shoulder height, subject's height and repetition. The explanatory power of such regression model was 79.9% which is only slightly higher than the explanatory power of the model for the female subjects (78.6%) and nearly 25% higher than that for the whole sample. However, the subject's height and the repetition did not have a significant relationship with the average value of MHT, and the programme also found that the anthropometric characteristics of the male subjects were strongly correlated amongst them.

5.4 Discussion

5.4.1 The repeatability of the maximum holding time

The basic aim of the trials reported here was to determine the repeatability of the maximum holding time (MHT), which has been taken to measure the individual's endurance capacity to the loads imposed on the muscles of the shoulder by the abduction of both arms at three angles: 30°, 60° and 90°. These experimental conditions were designed to impose significantly different biomechanical loads on the shoulder. This was evidently achieved, as the significant difference between the abduction moments acting on the shoulders showed.

Although the maximum holding time varied considerably between individuals, and even within individuals, for the sample as a whole it remained fairly consistent throughout the repeated trials at each abduction angle. In fact, even though the average MHT increased from one trial to the next, such increase was far from statistical significance, as shown by the results of analysis of variance. So, it was evident that - as a group feature- the maximum holding time was highly repeatable for the sample of subjects involved in the present study. The same conclusion emerged from the analysis on the length of MHT achieved by the subjects who participated in the first experimental stage of this study. Indeed, these results are in total agreement with the statements by Dul et al (1990, 1991), who reviewed a series of studies of endurance to isometric contraction.

5.4.2 Main influences on the MHT for postural loading

The average maximum holding time of the male subjects was significantly larger than that of the female subjects at each of the three abduction angles studied. Also, the increase in the abduction angle (which was shown to provoke an increase of the biomechanical loading on the shoulder) caused a significant reduction in the MHT. When calculated in relation to their longest MHT (that achieved at 30°), such reduction was, on average, approximately of the same magnitude for male and female subjects. The increase of the abduction angle also reduced the variability of the maximum holding time, and this effect was more noticeable among the female subjects.

The reduction of the MHT with the increase of the abduction angle may be explained by the increase of the abduction moment that came with it. This factor also had the effect of reducing the gap between the subjects with the longest holding time and those with the shortest one, as evidenced by the reduction of the coefficient of variation. That the male subjects achieved larger MHT than the females at the three abduction angles is a result that might be explained by attributing it to the difference in absolute muscular strength between the two gender groups (McArdle et al, 1991, p 457). It is also quite appealing to think that it might be explained by the presence of a significantly larger proportion of slow-twitch fibres in the muscles of the male subjects, but there is little evidence that such is the case (McArdle et al, 1991, p360). However, the finding of a larger endurance time for the male subjects differs from the results reported by Takala et al (1993), who observed that the endurance of males and females was not significantly different when they had to hold the right arm extended to the front with a weight suspended from the wrist.

In summary, the length of the MHT was significantly reduced by the increase of the abduction angle, which also had the effect of reducing its variability. Besides, the MHT was significantly larger for the male subjects than it was for the females, and it was not affected by the repetition of the trials.

5.4.3 A model to explain the variation of the maximum holding time

Given the wide inter-subjects variability of the maximum holding time found throughout the trials that formed the experiment here reported, it was deemed important to try and find a model to explain as much of it as possible. Besides looking for a model that included the data obtained from the whole sample of ten subjects, the search was also for models that considered separately the information collected during the trials performed by male and female subjects.

The variables to which the variation in MHT might be attributed were connected either to the experimental design or to the subjects themselves. The first category was constituted by the trial order and the abduction angle, which in turn determined the size of the moment acting on the shoulder joint. The second group had the weight and height of the subject, the shoulder height, the length of the arm and the length of the forearm. It was assumed that the relationship between MHT and those variables was most likely of linear nature.

The regression model with the largest explanatory power for the whole sample (56.5%) included seven variables. However, those that showed to have a significant relationship with the average value of MHT for the whole sample were (their specific explanatory power shown between brackets) the abduction angle (31.5%), the height of the subject (16.2%), the shoulder height (4.0%), the arm length (3.1%) and the subject's weight (1.4%).

The regression models for the sub-samples of male and female subjects had a considerably larger explanatory power. The one for the females explained 78.6% of the variation in MHT and included 8 variables. Those that were significantly related to the average value of MHT were the abduction angle (42.5%), the shoulder height (26.8%), the subject's height (5.9%), and the abduction moment acting on the right shoulder (3.0%).

For the results obtained from the trials with the male subjects, the best regression model explained 79.9% of the variation in MHT and included seven variables. The most significant relationship with the average value of MHT was for the variables abduction angle (42.8%), height (14.2%), shoulder height (12.7%), arm length (4.0%), and weight (3.8%).

Thus, for both male and female subjects it was possible to find a regression model that explained around 80% of the variation in MHT. Also for both groups it was the abduction angle the individual variable with the largest explanatory power, it was practically the same in both cases (43%) and it alone represented half of the total explained by the model. The next best variables for the group of female subjects were the shoulder height and the stature, whilst for the male subjects it was the stature and the shoulder height. However, since stature and shoulder height had an extremely strong correlation ($r= 0.976$ for the female subjects, $r= 0.980$ for the males) their added explanatory power may be attributed to stature alone.

To sum up, the variation in the length of MHT may be adequately explained by a linear regression model. The proportion of variation explained when the groups of male and female subjects were considered separately was approximately 80%, but that proportion went down to around 60% when the results were pooled together. Either way, most of the explanatory power of the model (nine-tenths of the total, at least) rests on just two variables: the abduction angle and the subject's height.

However, it is convenient to draw attention to the fact that the search for a model was above all an effort to identify the factors more likely to explain the variations observed in the maximum holding time for postural loads. It was not intended for the regression model to be used as a predictive tool for the calculation of that time and therefore not a great deal of effort went into refining its mathematical structure, which to the trained eye might even appear clumsy, particularly with respect to the balancing of units. The next chapter will deal precisely with the search for a model which may be used to predict the length of the maximum holding time, only this will be in function of the perception by the subjects of the fatigue process that occurred during the holding of the postures.

5.5 Conclusions

The iterative measurement of the maximum holding time for postural loading was carried out with the purpose of fulfilling the second and third specific objectives set to the present investigation, namely, to find out how repeatable is the MHT - which is assumed to be indicative of the endurance to postural effort- and to determine how that measure is affected by the changes in the abduction angle and by the gender of the subject.

Having analysed the results of the experimental work, three conclusions can be expressed:

1) The maximum holding time was proven to be, as a group feature, a highly repeatable measure of the endurance to postural loads. Nevertheless, the actual MHT achieved by the individual subjects during the repeated measurements evidenced substantial inter- and intra-subject variability;

2) The increase of the abduction angle had the effect of reducing significantly both the length of the maximum holding time and the dispersion of the individual measurements;

3) The maximum holding time for postural loading was significantly larger for male subjects than it was for females, although a considerable degree of overlap can be expected.

CHAPTER 6

RELATIONSHIP BETWEEN THE MAXIMUM HOLDING TIME AND THE PERCEPTION OF DISCOMFORT

6.1 Introduction

This chapter deals with the information provided by the subjects as to how they perceived the discomfort to grow whilst holding the posture randomly assigned in each of their nine trials. This information was collected in search for a better understanding, first, of the relationship between the passage of the holding time and the subjective perceptions of discomfort, second, of how that relationship might be affected by the change in the abduction angle and, third, of whether the nature of the relationship is the same for males and females. Those are the goals enclosed by the fourth specific objective of this investigation.

6.2 Experimental procedure

Borg's 10-point category-ratio scale (Borg, 1982; illustrated in figure 3.7) was used to obtain ratings of discomfort from the subjects whilst they held the postures. The subjects returned ratings for the 11 body regions depicted in a body mapping, which is shown in figure 3.6. The procedure followed in the collection of the discomfort ratings was described in chapter 3.

In this second series of trials, the quality of the information gained depended heavily on ensuring that the subjects were well familiar with each of the 11 regions shown in the body map, so that they could return a discomfort rating precisely for the region whose identifier was being called out by the experimenter. Thus, before they started the first holding, the subjects were shown on their own body where to find the boundaries between the body regions, and were instructed to think only of the region for which they were returning a rating. A rehearsal followed this explanation, and the procedure was repeated as many times as needed, until the subject was satisfied that they could clearly identify only the region being called out. Also, before they started each trial, the subjects were reminded that as soon as they felt that discomfort in any part of the body had reached the maximal intensity they could possibly bear (a rating beyond 10) they should call the effort to its end, and that they should try and be as consistent as possible in identifying such extreme sensation in every trial.

Thus, discomfort ratings were obtained at the beginning of each trial, and from then on every 60 seconds (120 seconds at some stages of the longest trials). The last rating was collected at the moment the subjects informed the experimenter that they were about to stop the effort due to unbearable discomfort.

6.3 Results

6.3.1 Background

A total of 90 trials was planned: 3 at each of the 3 abduction angles by each of the 10 subjects. However, in the middle of the third trial at 30° by subject No. 6 the equipment used for the collection of EMG signals presented a substantial failure which forced the experimenter to call off the trial. Both experimenter and subject were prepared to run the trial again, but the latter became unavailable due to the proximity of the final examinations; therefore, only the results from 89 trials will be presented. The data pool described and analysed in this chapter consisted of a total of 1495 discomfort ratings and the corresponding sampling times; of these, 726 were obtained from trials at 30°, 505 from trials at 60°, and 264 from trials at 90°. Breaking down the data pool by subject, 60 were obtained from subject No 1., 130 from subject No. 2, 172 from subject No. 3, 80 from subject No. 4, 107 from subject No. 5, 196 from subject No. 6, 108 from subject No. 7, 98 from subject No. 8, 278 from subject No. 9 and 266 from subject No. 10.

Of the 89 trials successfully completed, on only three occasions was the holding stopped without the subject returning a rating of 'maximal' for the discomfort they were experiencing. All three instances involved holdings at 30° by the male subjects with the largest MHT. In one case, it was the subject himself who asked for the trial to stop because, although he was experiencing the worst discomfort in the arms (his ratings were in fact 10 at that stage), he suddenly felt he could not stand any longer and feared he could fall over. Afterwards, he explained that it was not a case of dizziness, but he felt his knees were locked and he could not manage even a slight shift on his feet.

In the other two instances, the subjects had apparently reached a ceiling. Although their discomfort was evident (heavy breathing, grinding of teeth, red face) they still struggled trying to keep the posture for longer, as if reluctant to give up the effort. Since this clearly contravened the instructions issued to all the subjects in the sense of avoiding the prolongation of the effort just for the sake of it, the experimenter took the decision to finish the trial. When the subjects were told afterwards that they appeared to have tried to push themselves too hard, they agreed with this appreciation.

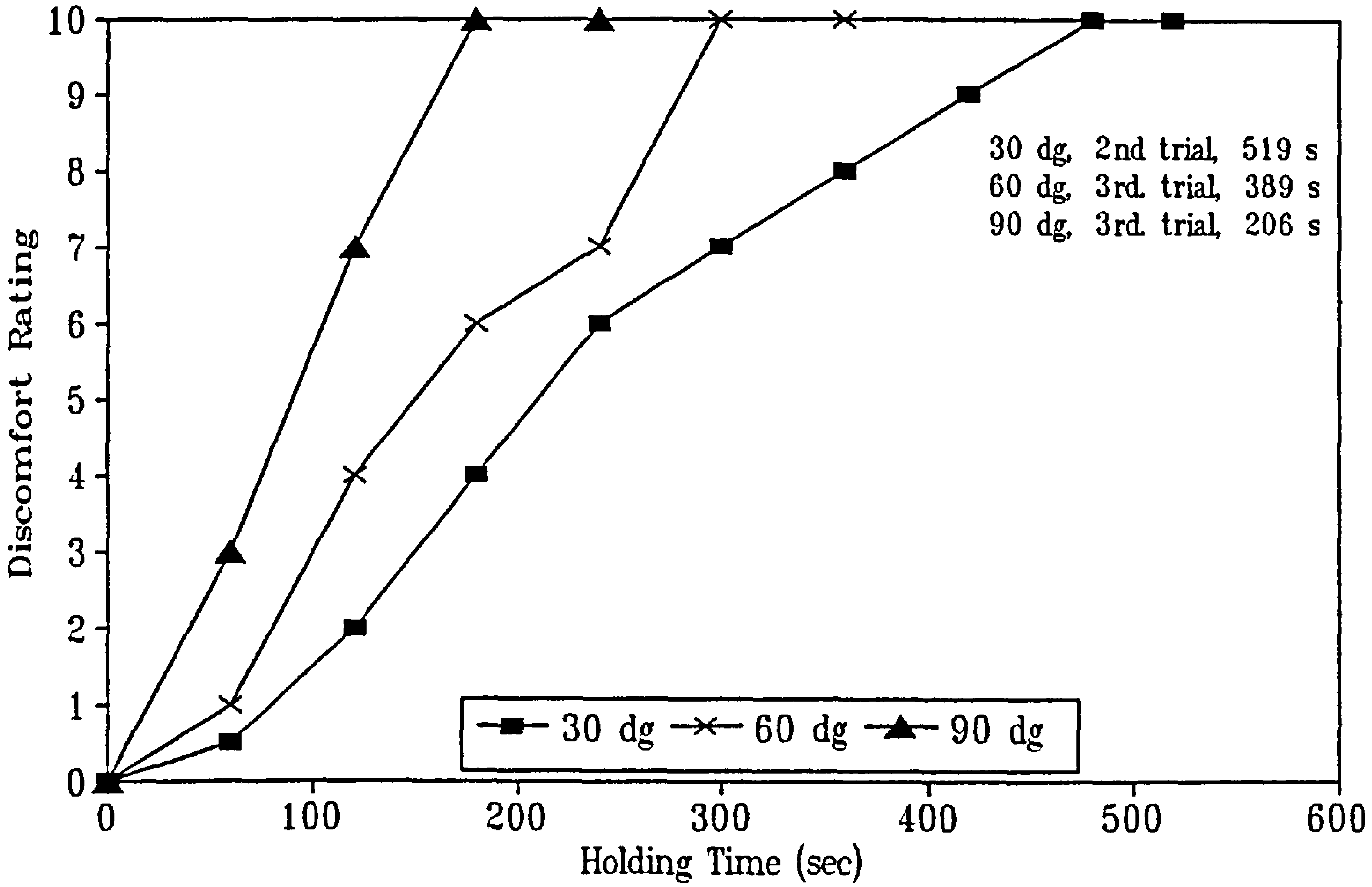
6.3.2 Sites of unbearable discomfort

In all the 89 trials, the regions corresponding to the deltoid muscles were the ones accorded the highest rating of discomfort, with 'maximal' on 86 occasions, 10 on the remaining three. No other region ever reached that level of discomfort. The results can be summarised as follows: 60 times the subjects reported the worst discomfort as affecting equally the right and left medial deltoid, 16 times it affected mainly the right medial deltoid, 8 times it was the left medial deltoid the worst affected, and 6 times the medial and posterior deltoid muscles on both arms were equally affected.

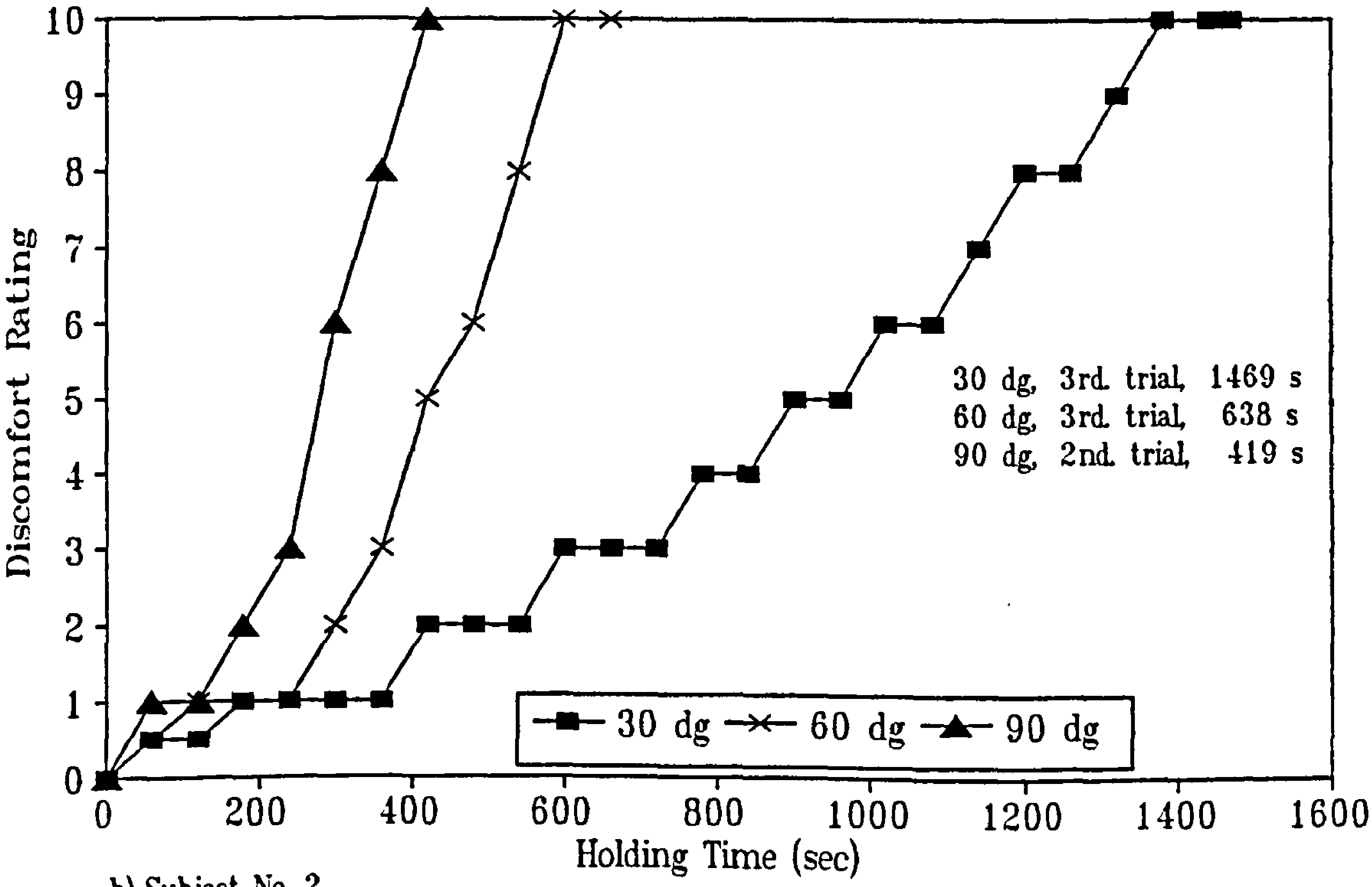
Since the medial deltoid muscle was identified as a site of worst discomfort in every one of the 89 trials, the discomfort ratings returned for this muscle were the ones considered in all further analysis. Therefore, in the remainder of this thesis, references to 'discomfort rating (s)' actually stand for 'the discomfort rating (s) returned for either of the medial deltoid muscles'.

6.3.3 The time course of the discomfort ratings

As demonstrated in chapter 5, the maximum holding time of each subject to each of the three abduction angles investigated did not change significantly between trials. Therefore, the course of the discomfort ratings over the holding time is presented here for one trial at each angle, which is representative of what happened during the other two trials. In each case the trial with the longest holding time was chosen and these are shown in figure 6.1 (a) - (j). The values plotted on the horizontal axis of these figures represent the actual duration (in seconds) of the trials, and the actual holding times for the three abduction angles are also recorded in the key to each graph. It is convenient to mention that although the actual holding times varied from 206 seconds for subject No. 1 at 90° to 3623 seconds for subject No. 9 at 30°, the length of the horizontal axis in figure 6.1 is the same for all the graphs. This means that the scale on this axis is not uniform, it depends on the length of the trials being represented, and in all the cases it extends from 0 seconds to the duration of the longest trial in that particular trio.

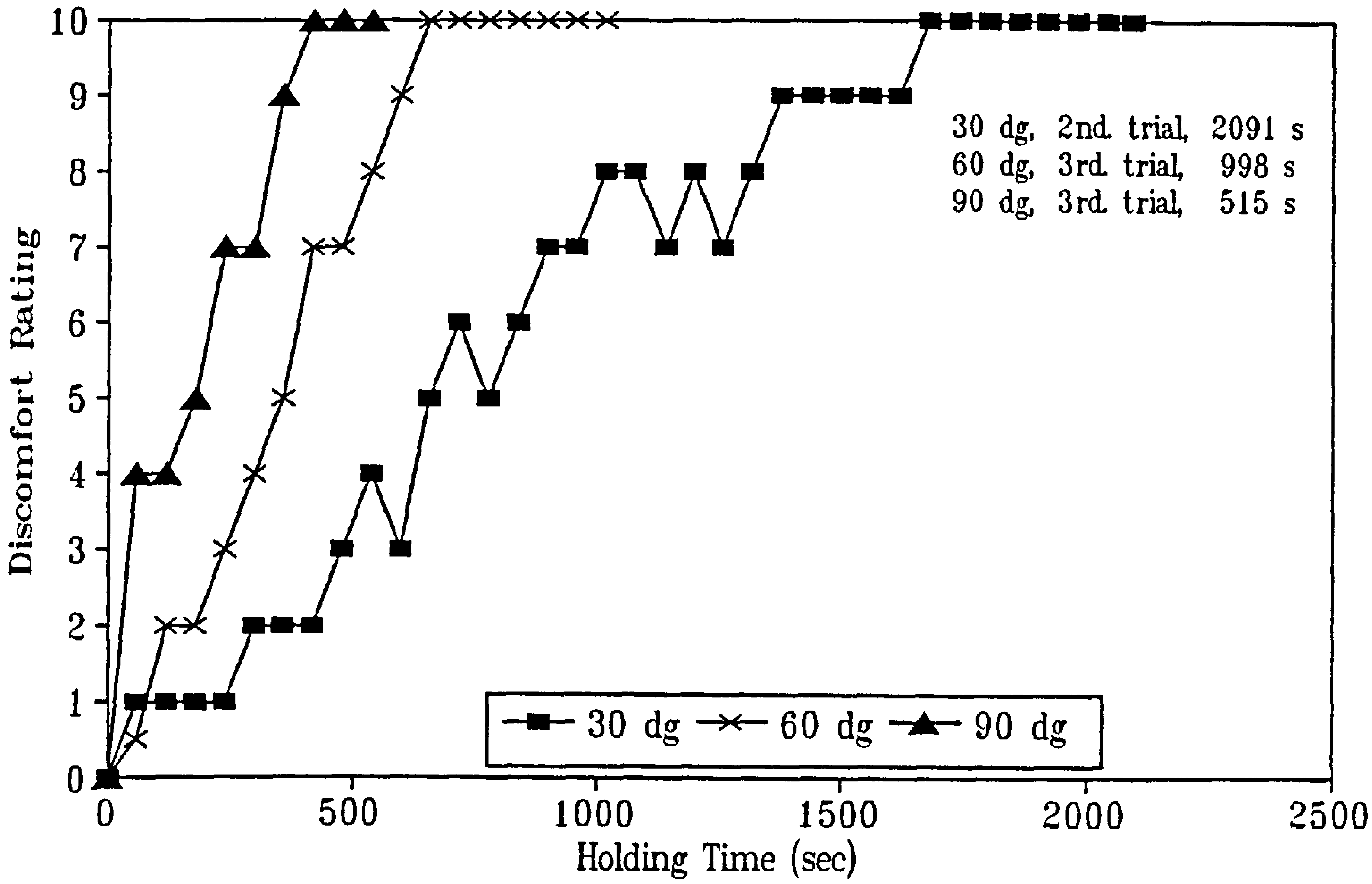


a) Subject No.1

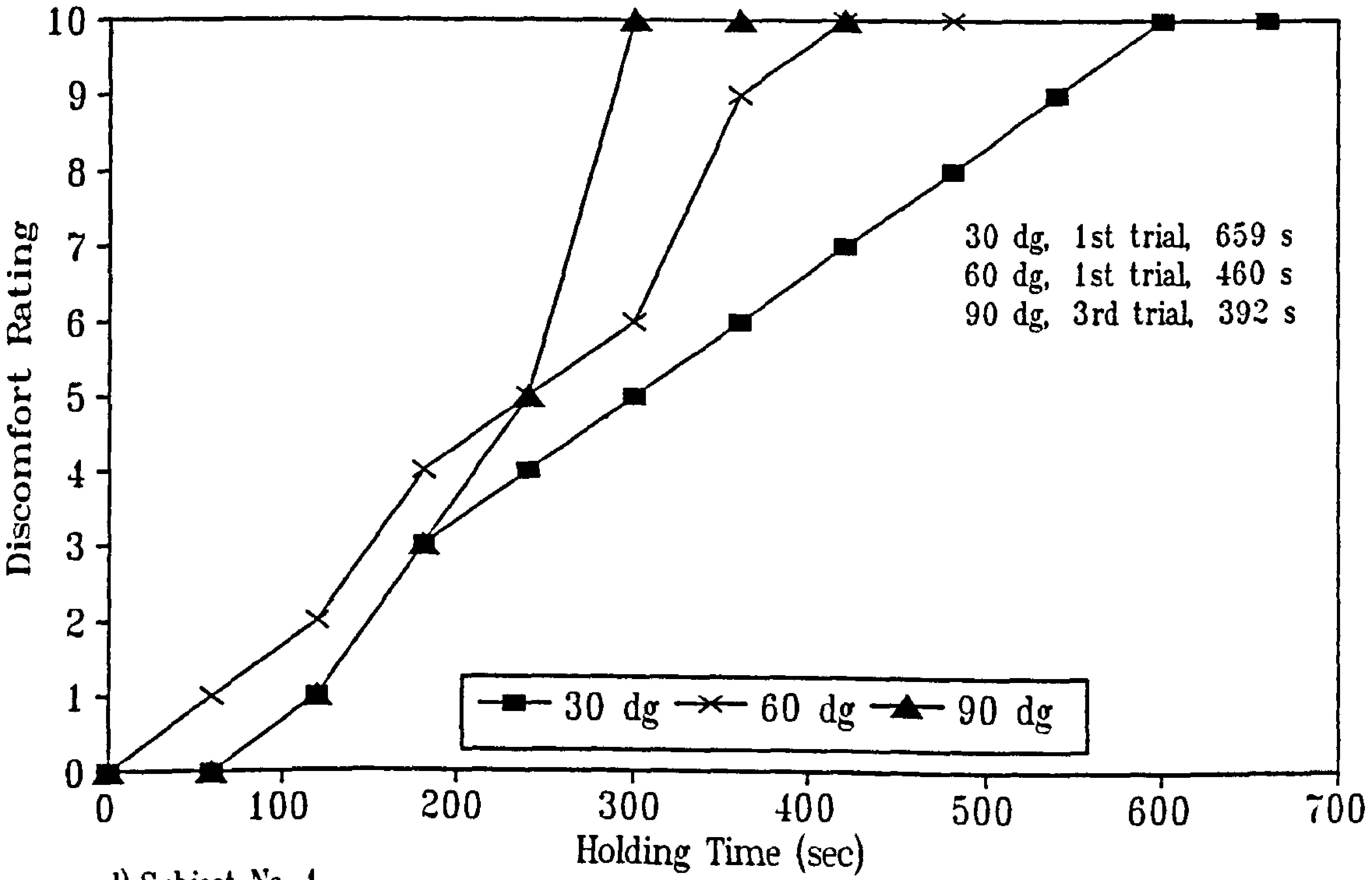


b) Subject No. 2

Fig. 6.1 Discomfort ratings returned by each subject during their longest trial at each angle.

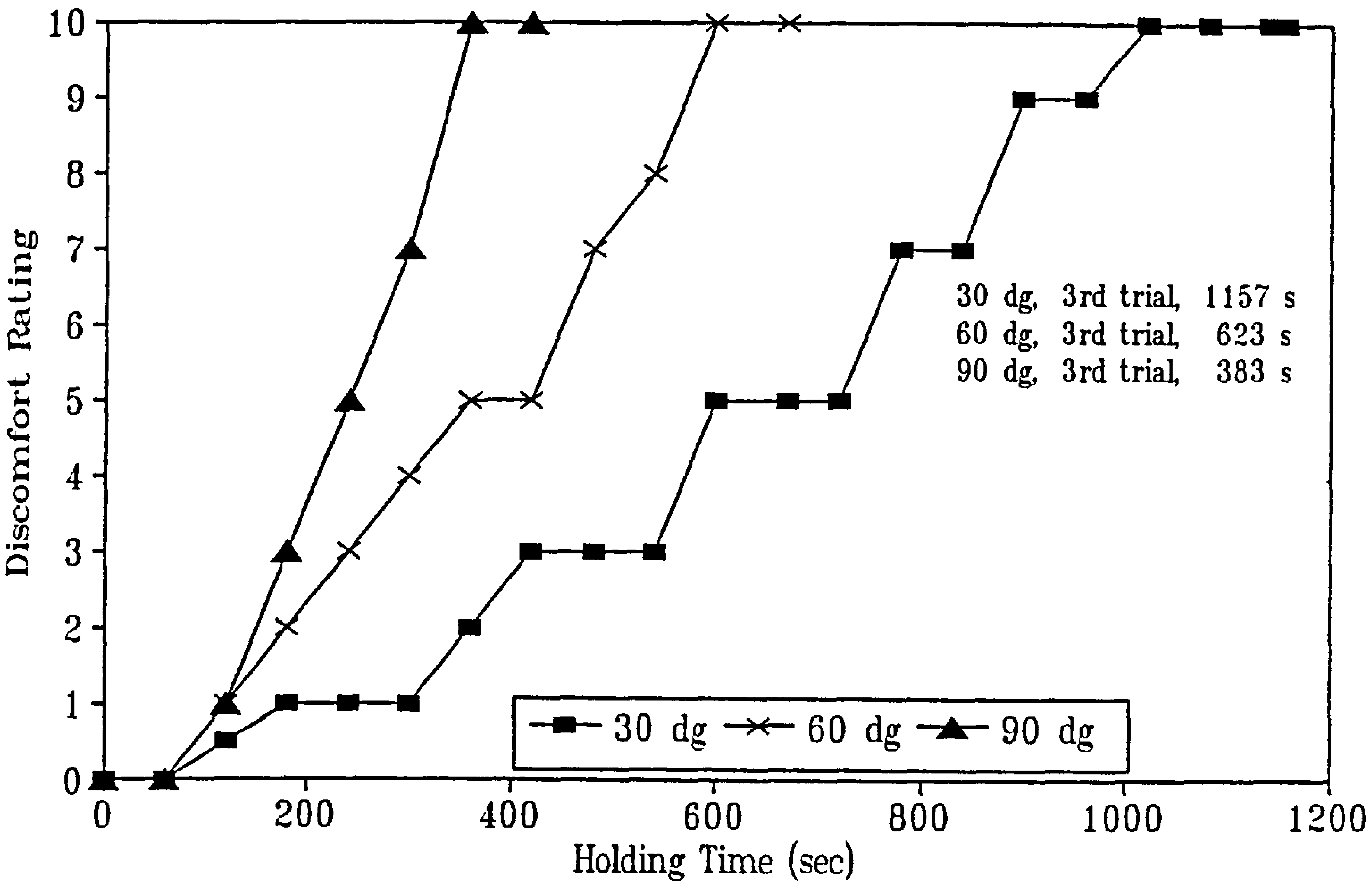


c) Subject No. 3

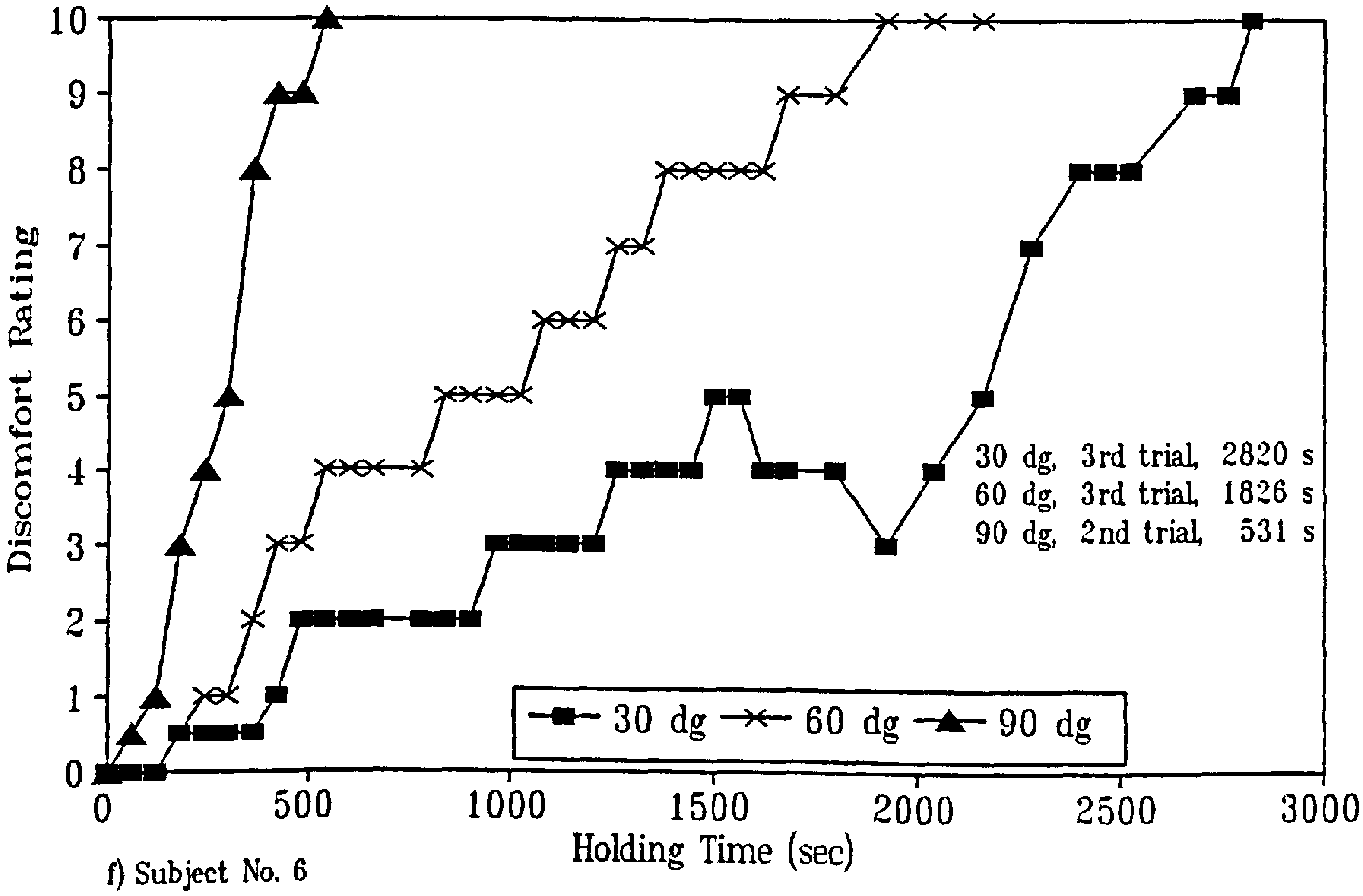


d) Subject No. 4

Figure 6.1. Continued

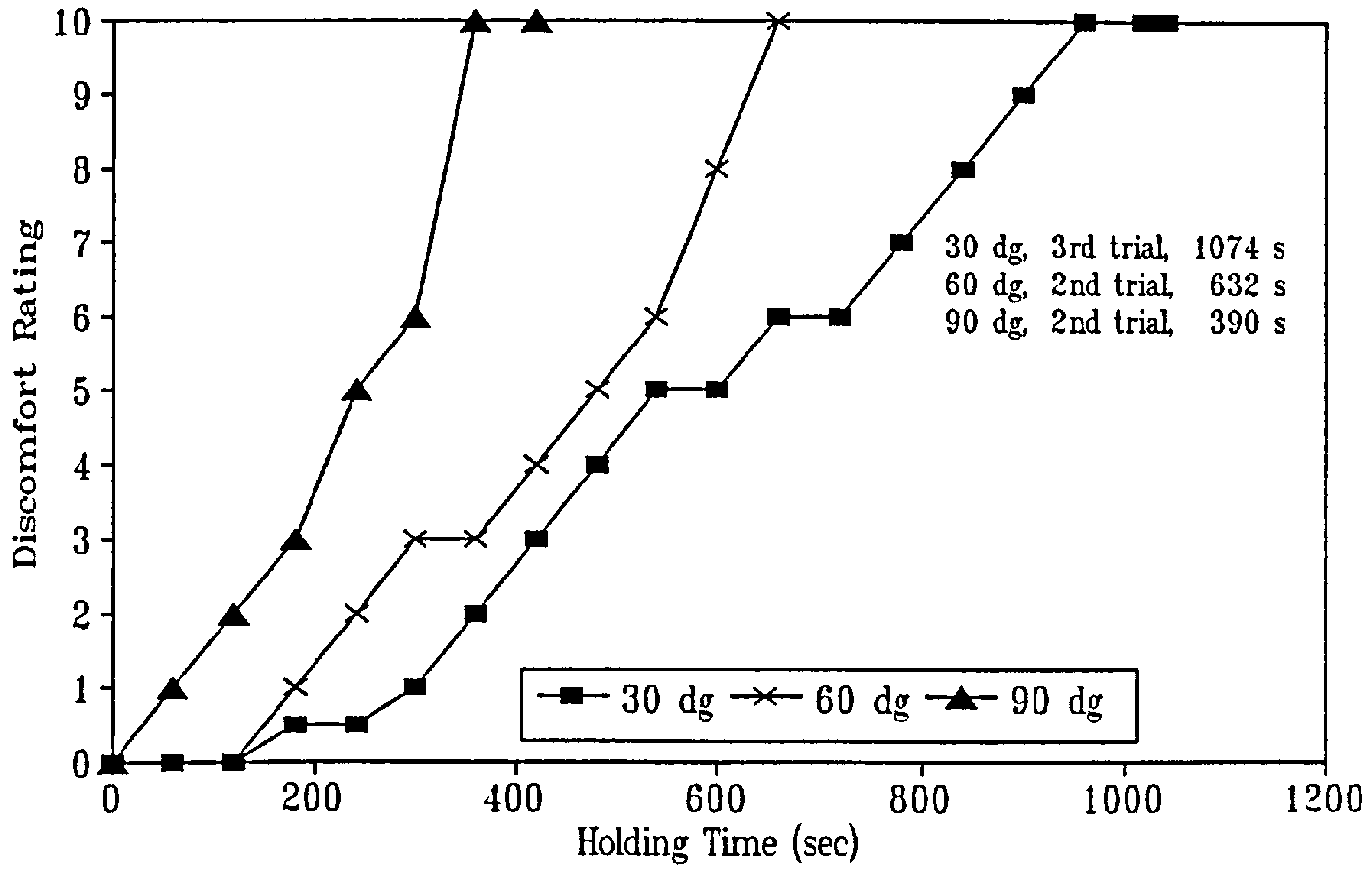


e) Subject No. 5

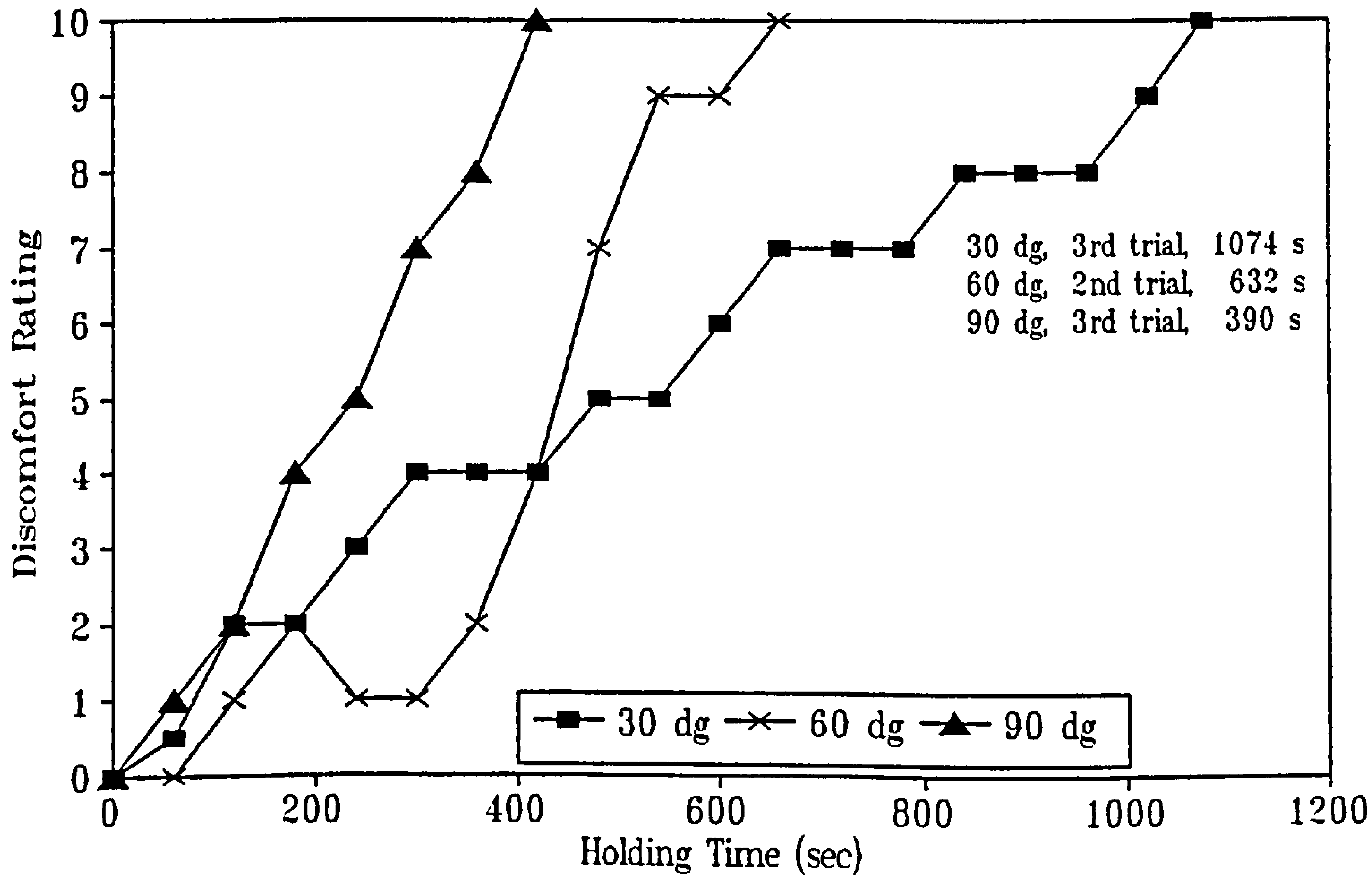


f) Subject No. 6

Figure 6.1. Continued

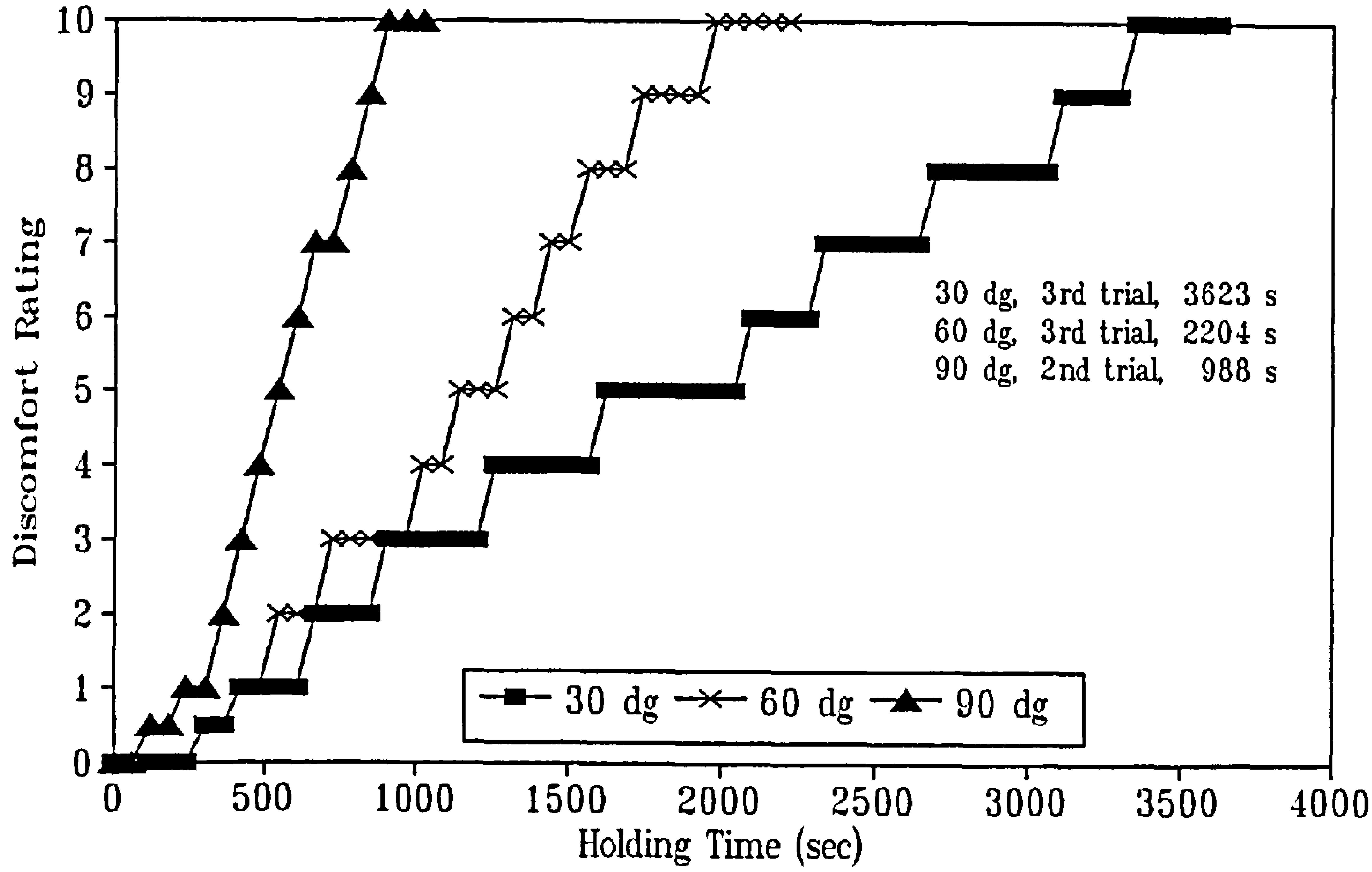


g) Subject No. 7

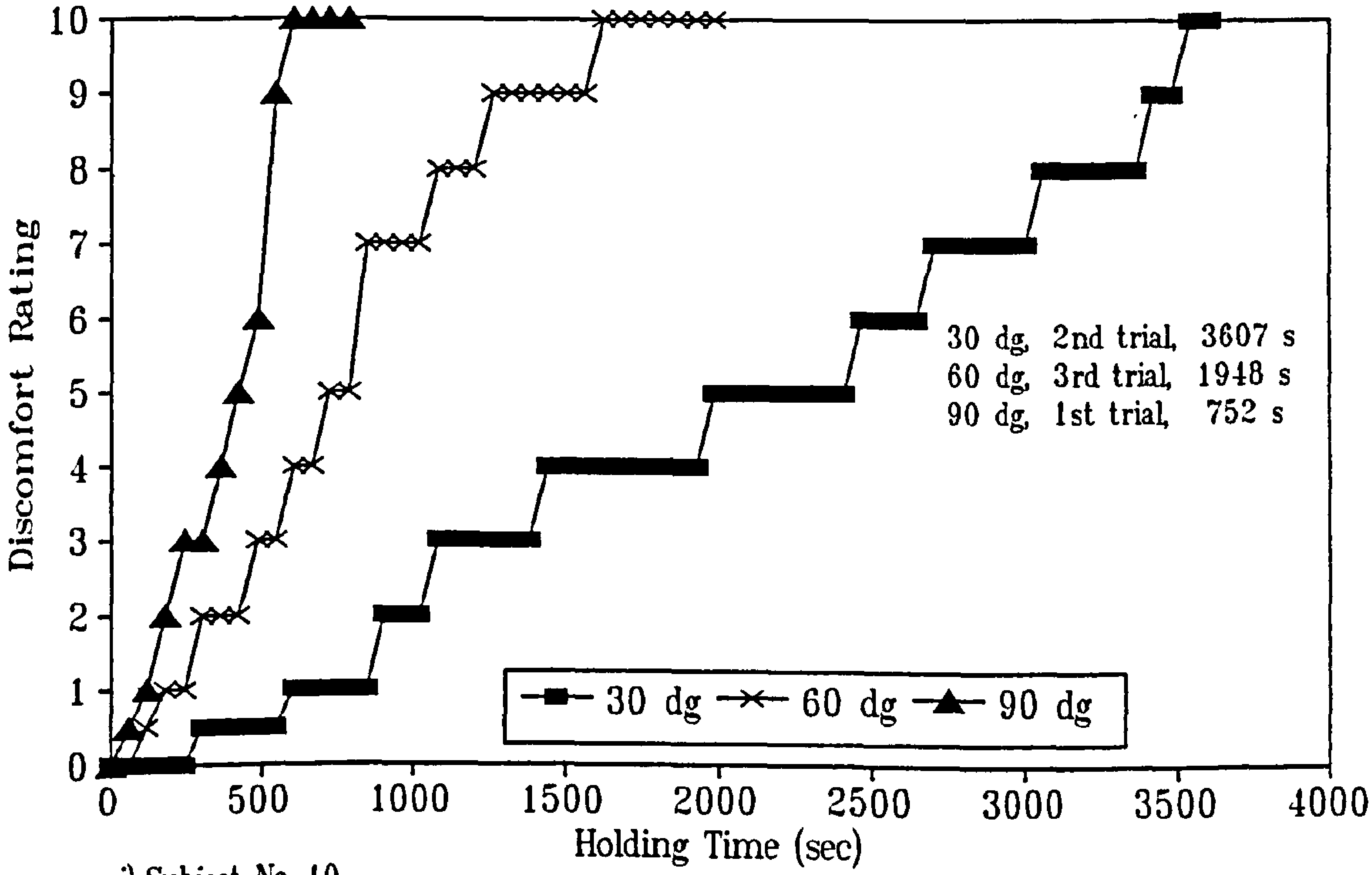


h) Subject No. 8

Figure 6.1. Continued



i) Subject No. 9



j) Subject No. 10

Figure 6.1. Continued

It is important to state that no numerical value was assigned to what the subjects perceived to be the 'maximal' level of discomfort, the one that forced them to stop the holding. This was so mainly because during the early trials, when asked to match a numeric value to the maximum level of discomfort, most of the subjects could not give a definite answer, but they could tell it was quite different from the sensation they had just rated as 10. They also said that often the passage from 10 to 'maximal' was quite sudden, and did not allow them to stop and think "How much is that?" before calling the holding to a halt.

This left the researcher with two options: either to keep the very last discomfort rating as a 10, or to arbitrarily choose by himself a higher value, with the risk of introducing yet another element of subjectivity into the rating process. The judgement was that the first option would make better sense, since it would reflect more closely the perception of discomfort by the subjects, right to the very end of the holding. Therefore, the plots presented in figure 6.1 (a) - (j) show the discomfort ratings on an axis with a scale from 0 to 10, with the last datum marked as 10, which effectively equated the maximal discomfort the subjects experienced at the moment they stopped the holding with a rating of 10.

Setting this limit on the scale of discomfort ratings could certainly affect the shape of the relationship between the holding time and the subjective perception of discomfort, and this issue will be dealt with in section 6.4, where the statistical procedures applied to the data will be described and discussed.

However, the absence of a "true maximal" is also an important methodological issue, and in this regard it will be considered at length in the Discussion chapter.

6.4 Discussion

6.4.1 Nature of the relationship between the holding time and the subjective perception of discomfort

The main aim behind the collection of discomfort ratings was to probe for the existence of a significant relationship between the passage of the holding time and the subject's perception of how their discomfort grew during it. The graphs plotted in figure 6.1 (a) - (j) show that, with only two exceptions (subject No. 6 at 30° and subject No. 8 at 60°), the growth of the discomfort ratings followed a fairly linear pattern, which was more evident as the abduction angle increased. Furthermore, it looked as if at each one of the three angles studied the pattern was similar for all the subjects. Therefore, it might be expected that a linear regression model would adequately express the relationship between the holding time and the discomfort ratings.

As seen in chapter 5, although the holding times for each subject at each abduction angle were fairly consistent, there was a considerable variability between subjects. Therefore, if the actual values of holding time were used to fit a regression line to the data of discomfort rating, it would be strongly affected by the variability between subjects. Besides, it would be necessary to fit a line separately to the data obtained at each abduction angle, to account for the difference in slope created by the shortening of the holding times that came with the increase of the abduction angle.

In order to assess in a consistent manner the growth of the subjective perception of discomfort in the course of the 89 trials, the sampling times during each individual trial - i.e., the time when a discomfort rating was returned- were normalised against the duration of that trial, so that they were converted into percentage of the maximum holding time. A scatter plot of the normalised data for all the trials is presented in figure 6.2. This graph shows that there was in fact an obviously linear pattern in the way the discomfort ratings grew over the holding time.

Separate graphs for the discomfort ratings returned during all the trials at each abduction angle were also prepared, as presented in figure 6.3 (a) - (c). The assumption about the pattern of change being similar for the three angles when the times were normalised appeared to be correct, although at 90° the data were more widely spread than at the other two angles. This issue will be considered again in section 6.4.2, when the pattern of change is turned into an equation to express the relationship between holding time and discomfort ratings.

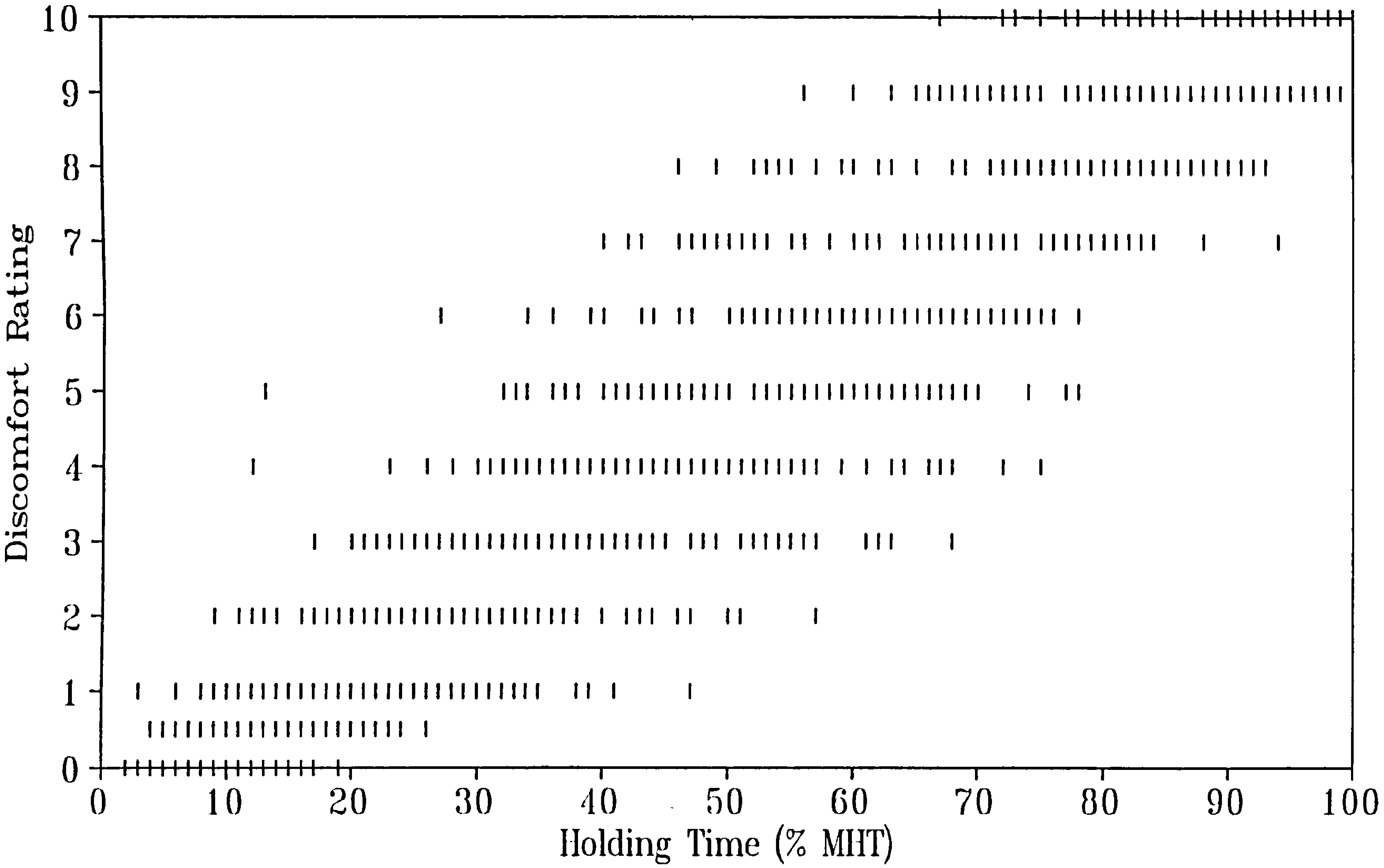
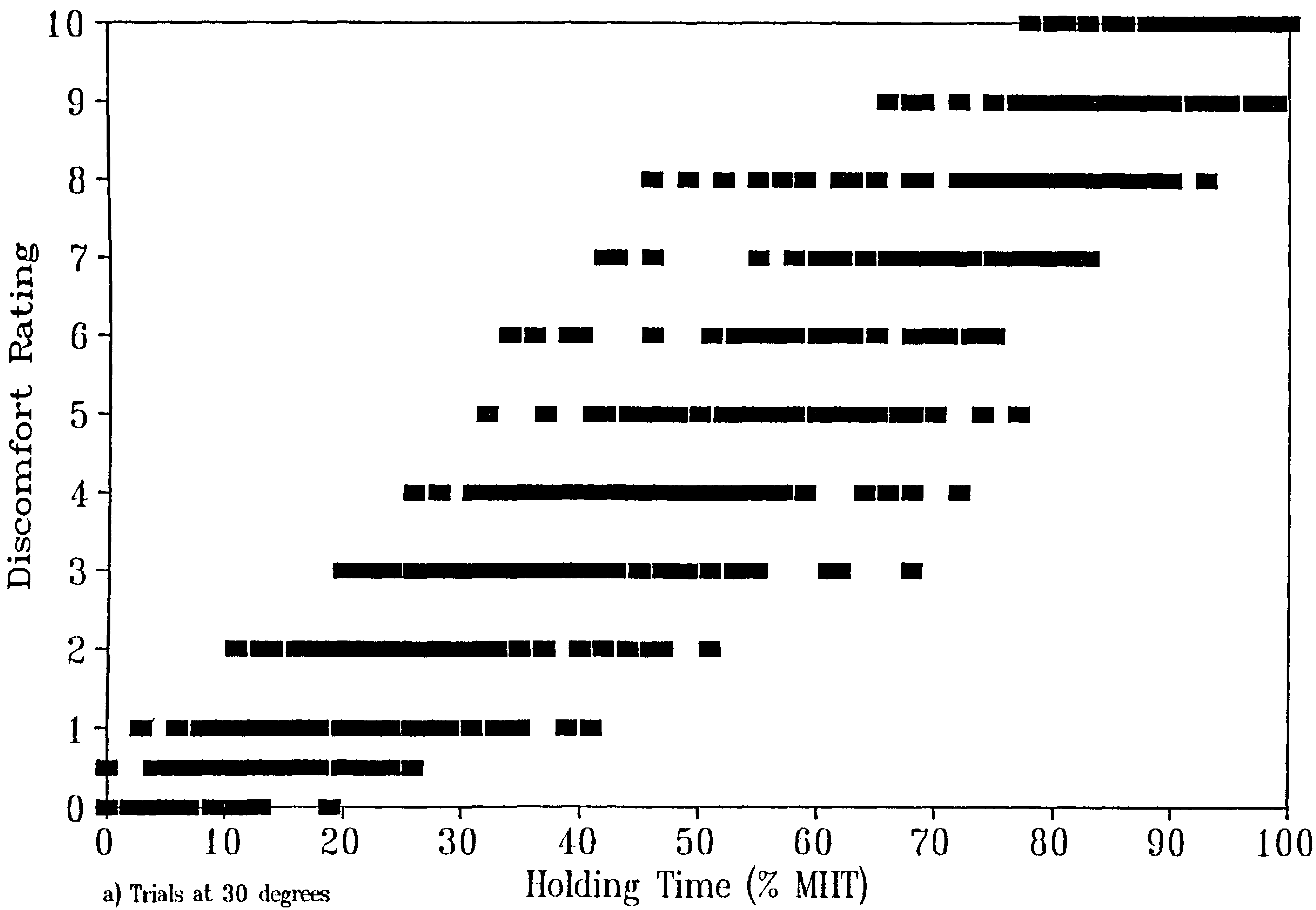
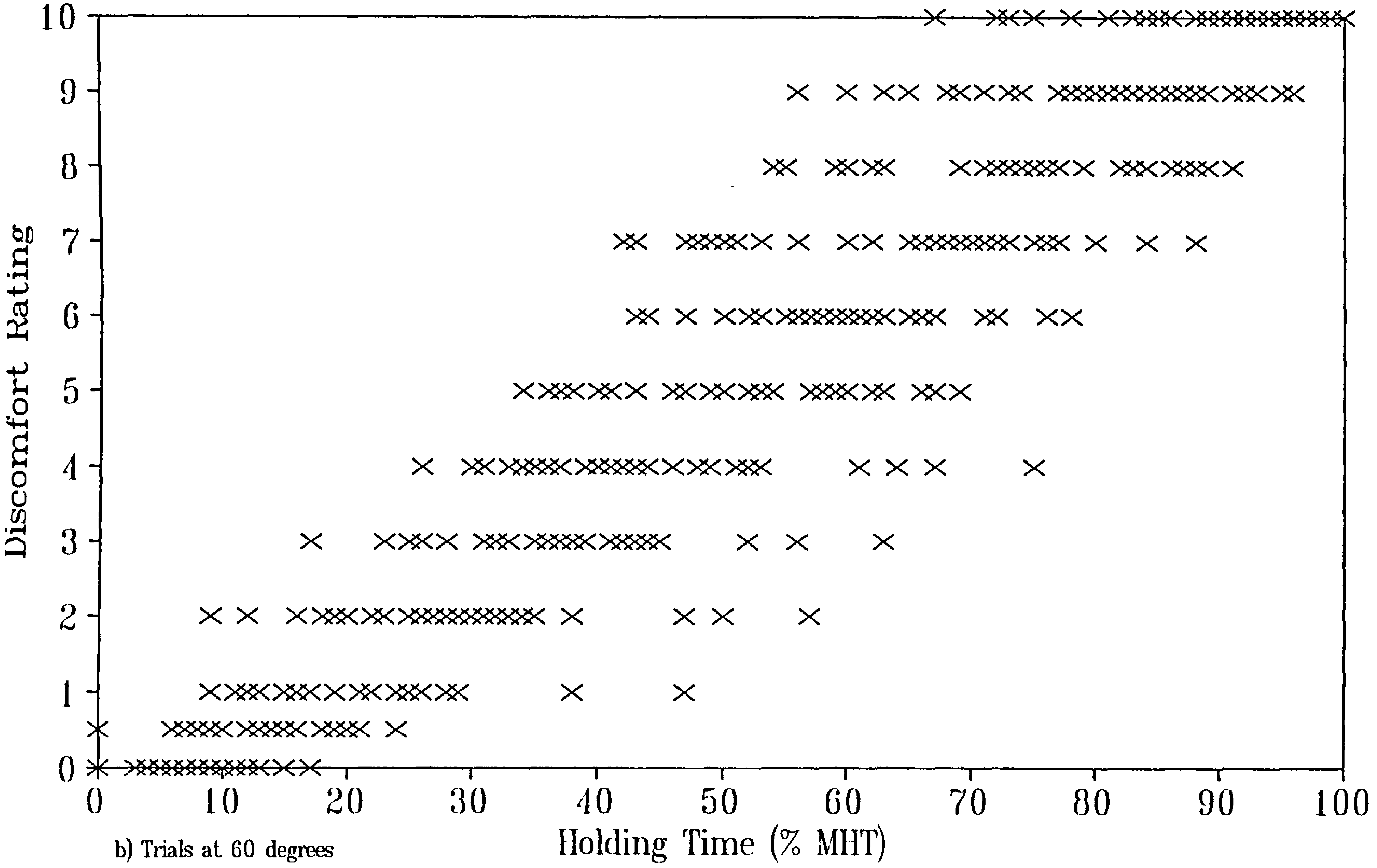


Fig. 6.2 Scatter plot of the 1495 discomfort ratings collected during 89 trials to measure the maximum holding time to three shoulder-loading static postures. Each dash may represent more than one data point.



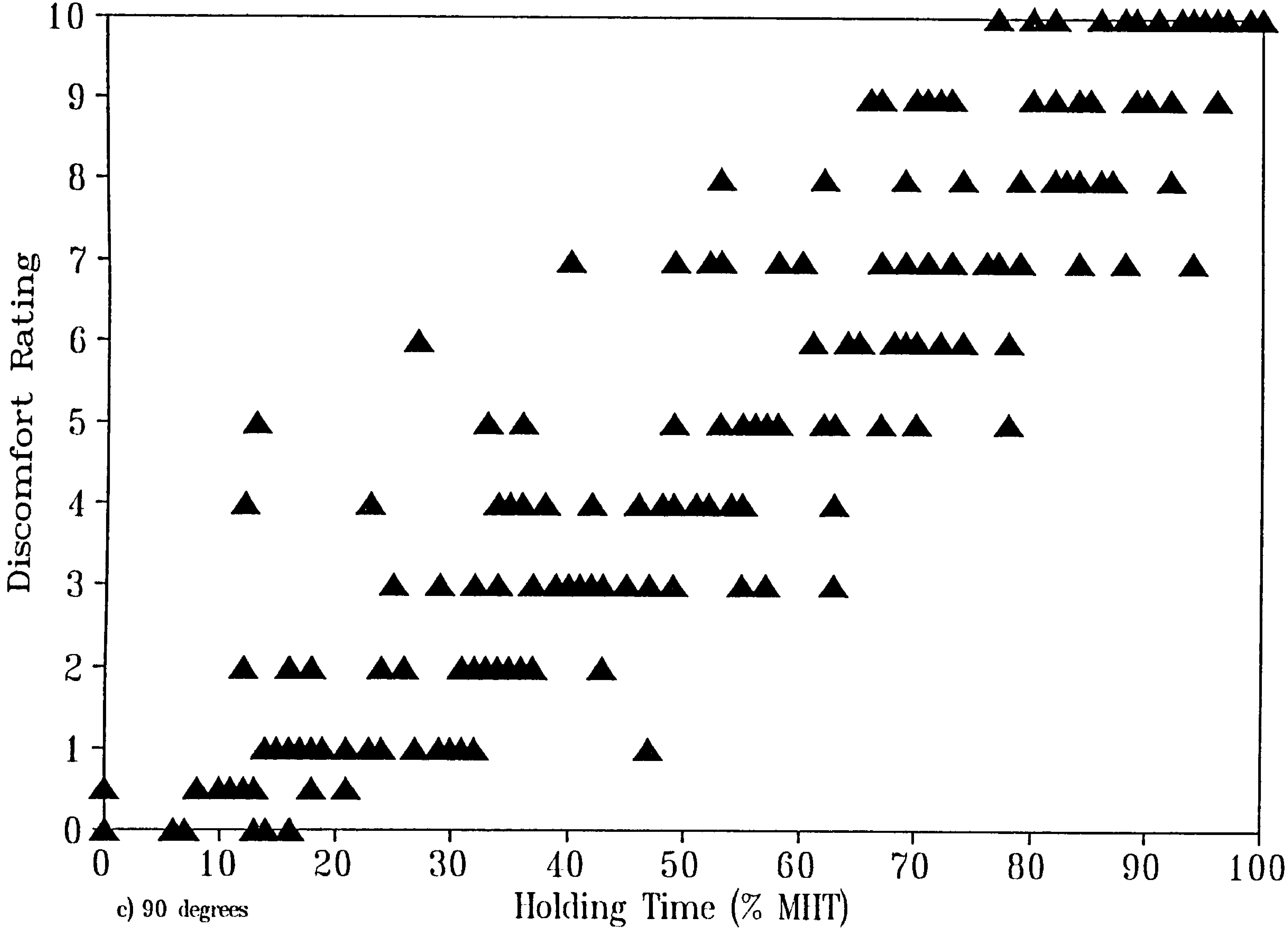
a) Trials at 30 degrees

Figure 6.3 Scatter plot of the discomfort ratings collected during the trials at each abduction angle



b) Trials at 60 degrees

Figure 6.3. Continued



c) 90 degrees

Figure 6.3. Continued

The shape of the scatter plot presented in figure 6.2 strongly suggests the existence of a linear relationship between the discomfort ratings and the percentage of the maximum holding time. A test of linear correlation on the 1495 data points produced a correlation coefficient of 0.956, which was highly significant ($p < 0.0001$). The correlation coefficient for the 1495 data points consisting of discomfort rating and the actual holding time (in seconds) was only 0.575, which although still statistically significant at the same level ($p<0.0001$) is indicative of a weaker relationship. This is further demonstration of the convenience of normalising the holding times into percentage of the duration of the trial.

Finding that there was an overall linear correlation (with $r = 0.956$) for the 1495 pairs of values between discomfort ratings and percentage holding time was an important step, but this led to the question of whether this relationship was equally true for the data collected during each one of the trials, despite the differences in the abduction angle and in the gender of the subjects. To answer that question, the correlation coefficient for each of the 89 sets of data was calculated, and this produced values that ranged from a low of 0.781 to a high of 0.996. Coefficients of this size indicate the presence of very strong linear correlation between discomfort ratings and percentage holding time at the level of individual subject. The 89 correlation coefficients may be found in Appendix B.

When it came to assessing the significance of the correlation, only the coefficient for the data from the first holding by subject No. 1 at 90° failed to reach significance, even though the actual value was quite high, $r= 0.948$, $p>0.05$. The failure to reach significance was probably due to the very short holding time, with just three data points recorded.

6.4.2 Expression of the relationship between discomfort ratings and the holding time

Once it was found that the relationship between percentage holding times and discomfort ratings was of linear nature, the next step was to look for the most adequate means of expressing it. To do this, a linear regression model was fitted to the complete set of data points (a total of 1495) collected during the 89 trials, which have been already plotted to produce figure 6.2. The regression equation for the best-fit model was

$$\text{Discomfort Rating} = -0.509 + 0.107\ [\%\ \text{MHT}]$$

with a standard error of the estimate equal to 1.018 and coefficient of determination $R^2 = 91.5\%$. The graphic illustration for this equation is presented in figure 6.4

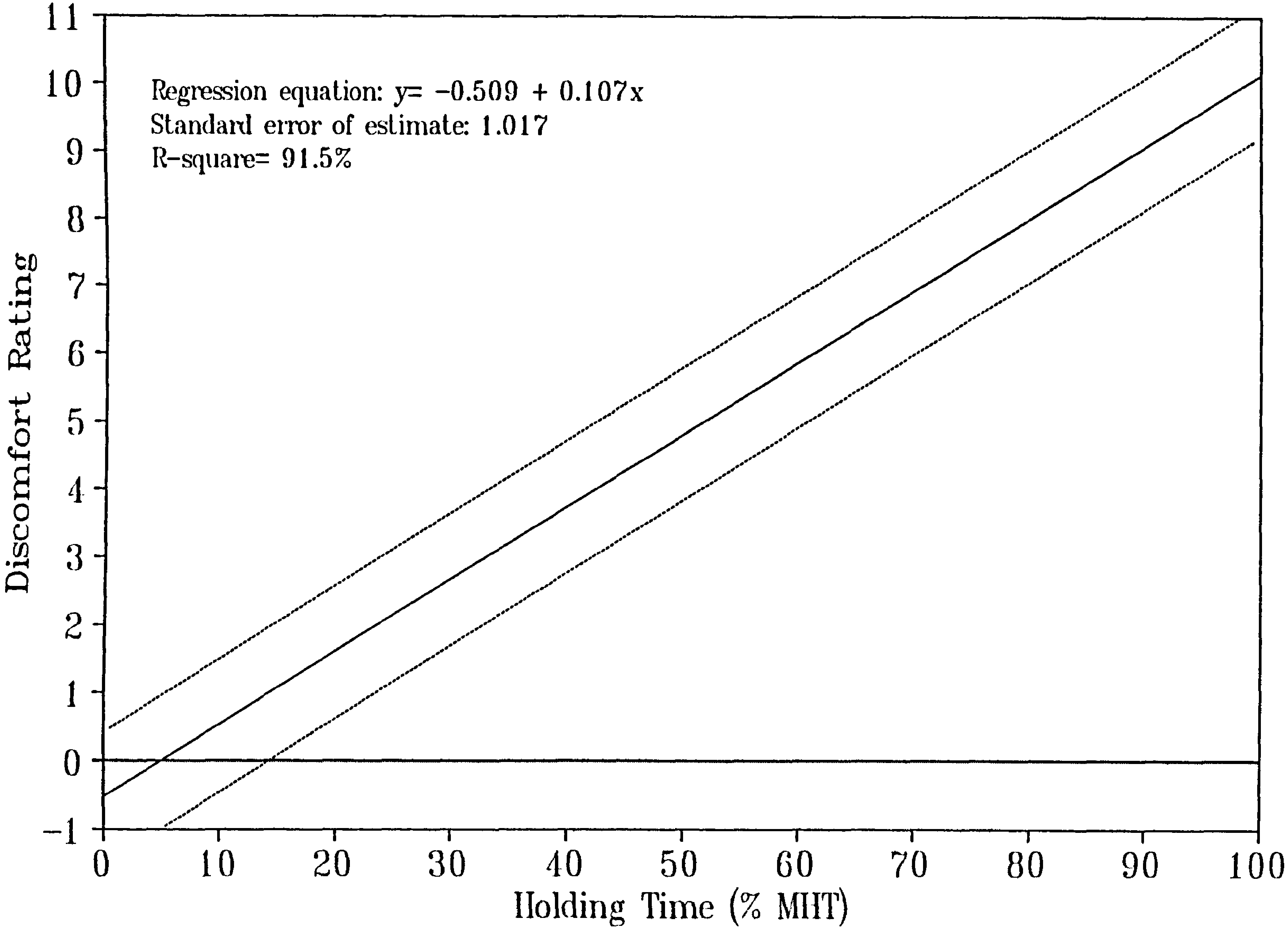


Fig. 6.4 Best-fit regression line for the sample of 1495 discomfort ratings on percentage holding time.
Dashed lines are traced at one standard error of the estimate.

In 85 out of the 89 trials reported here, at time 0 the subject returned a discomfort rating of '0' for both medial deltoid muscles. The initial rating returned for either of those muscles in the other four trials was '0.5'. Since most of the individual data sets started at (0,0) it is justified to fit a regression line with a forced intercept through the origin (Neter and Wasserman, 1974, pp 156-159). The equation for such regression line was:

$$\text{Discomfort Rating} = 0.0994\ [\%\ \text{MHT}]$$

the standard error of the estimate was 1.052 and the coefficient of determination, R^2, was 96.9%. This regression line is presented graphically in figure 6.5.

The appropriateness of fitting a regression line which goes through the origin is demonstrated with two statistical arguments. First, that the standard error of the estimate remains practically unchanged by the modification of the line's slope. Indeed, the displacement of the regression line meant a mere 3% increase in the variance of the dependent variable (discomfort ratings). Second, that the value of R^2, the coefficient of determination, increased (albeit rather moderately) to account for slightly more than an extra 5% of that variance (from 91.5% to 96.9%)

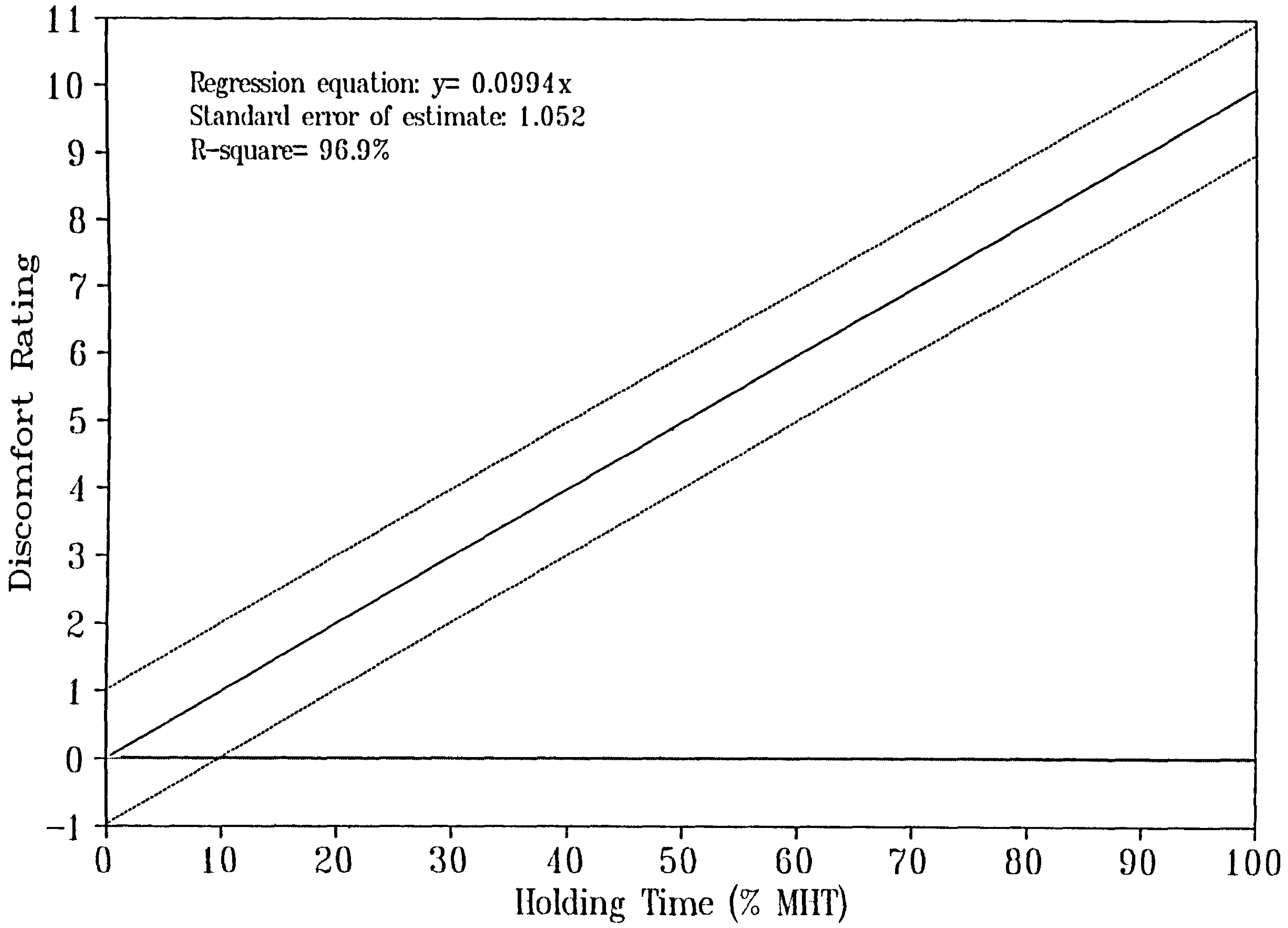


Fig 6.5 Regression line for the sample of 1495 discomfort ratings on percentage holding time, with forced intercept at the origin. Dashed lines are traced at one standard error of the estimate.

6.4.3 Influence of the abduction angle and the gender of the subject on the perception of discomfort

As mentioned in the introduction to this chapter, the objective set to the work now being described was threefold. The first part of that objective has been fulfilled by finding that, for the whole body of data collected during the experimental procedures, there is a linear relationship between the passage of the holding time and the growth of the perceived discomfort, and that a regression model accounts for well above 90% of the variance of the latter .

6.4.3.1 Effects of the increase of the abduction angle

To fulfil the second part of the objective, it is necessary to establish whether those features of the relationship between discomfort ratings and holding time appear the same when the data collected during the trials at each of the three abduction angles are analysed separately. In fact, this second subsidiary goal has been partially achieved already, since it has been shown in section 6.4.1 that the linearity of the relationship is present in all three subsets of data, and this was illustrated in figure 6.3 a) - c).

In order to complete the achievement of the second sub-objective, a separate linear regression model (with forced intercept at the origin) was fitted to the data collected at each abduction angle. The best-fit regression equation for each of those models was:

a) at 30°, Discomfort Rating = 0.0985 [% MHT],

with standard error of the estimate equal to 0.998 and R^2 equal to 91.2%;

b) at 60°, Discomfort Rating = 0.1016 [% MHT],

with standard error of the estimate equal to 1.057 and R^2 equal to 91.1%;

c) at 90°, Discomfort Rating = 0.0975 [% MHT],

with standard error of the estimate equal to 1.158 and R^2 equal to 89.9%.

The slope of the regression line fitted through the data collected during the trials performed at each of the three abduction angles remained quite close to that of the overall data pool (0.0994), suggesting that, despite the increase of the abduction angle, the pattern of growth of perceived discomfort remained largely the same. Nevertheless, the increase of the abduction angle was accompanied by the widening of the spread of the data, which was evidenced by the increase in the standard error of the estimate and (particularly for the change from 60° to 90°) the decrease of R^2. The wider spread of the data is clearly evident in the graphs presented in figure 6.3.

6.4.3.2 Comparison between the subjective reactions of male and female subjects to postural loading

The third ancillary purpose of this inquiry into the growth of perceived discomfort during the performance of a purely postural effort was to find out whether the gender of the subject could have a significant influence on the nature of the relationship between discomfort ratings and holding time. This goal was tackled by finding the best-fit regression model for the data collected during the trials performed by the individuals of each gender.

The regression equations generated by such procedure were:

1) For male subjects, Discomfort Rating = 0.0970 [% MHT], with standard error of the estimate equal to 0.917 and R^2 equal to 92.7%;

2) For female subjects, Discomfort Rating = 0.1030 [% MHT], with standard error of the estimate equal to 1.218 and R^2 equal to 88.6%;

According to this result, the gender of the subject appeared to have a stronger effect on the relationship between holding time and discomfort ratings than did the abduction angle. Whilst the slope of the regression line for the data collected from male subjects was only slightly lower than that of the overall regression line (0.0970 and 0.0994, respectively), the slope of the line for the data obtained from the female subjects was evidently larger ($0.1030 > 0.0994$). There were also important differences between the values of standard error of the estimate and of R^2, evidencing a much wider spread of the data for the female subjects than it was for the males.

6.4.3.3 Assessment of the effects of angle and gender considering the trials individually

Thus, a comparison of the regression equations fitted through the data collected during the 89 trials, grouped together in function of either the abduction angle or the gender of the subject, showed that whilst the increase of the angle appeared not to determine a difference in the subjective perception of the growth of discomfort, this was in fact different when the subjects were grouped according to their gender. To further test these impressions, a regression line for the data obtained from each of the 89 trials was calculated and their slopes submitted to analysis of variance, in order to assess the effects of the abduction angle, the gender of the subjects, and their possible interactions. The slope for each of the 89 regression lines may be found in Appendix B.

The results from the analysis of variance confirmed that whilst the relationship between holding time and discomfort rating was not different in function of the abduction angle, it was significantly affected by the gender of the subject. .The mean value of the slope for the regression lines fitted to the data obtained at 30° was 0.09862, at 60° it was 0.1000 and at 90° it was 0.0977; these values did not differ significantly between them, the ANOVA test reporting $F_{2,86} = 0.33$, $p > 0.7$. The mean value of slope for the regression lines for data from female subjects was 0.10189, and for male subjects was 0.0955. There was a significant difference between these values, with $F_{1,87} = 7.82$, $p < 0.01$.

To test whether the subjects perceived the growth of discomfort to be the same at the three abduction angles, irrespective of their gender, the slopes of the regression lines fitted to the results from the 89 trials were further analysed, applying the general linear model (GLM) facility of MINITAB, that allows to search for significance of each factor separately and of their interactions. This showed that the interaction between gender and abduction angle was not significant for either of the two groups of subjects, the F-value for the female subjects was 0.33 ($p>0.7$), and for the male subjects it was 0.01 ($p>0.99$).

These results suggest then that, on average, the subjects did in fact perceive the growth of discomfort to follow a very similar pattern at the three abduction angles studied, and this despite the significant differences in their maximum holding time for each condition, which were demonstrated in chapter 5. However, the gender of the subjects appeared to affect significantly their perception of the growth of discomfort, and this became manifest in two ways. First, the slope of the regression line for the female subjects was significantly higher than that for the male subjects, which might be interpreted as evidence that the latter tend to be more resilient to the sensation of discomfort provoked by the holding of the posture, although it might well be the case that they are simply more reluctant to report it. Second, the variance between subjects was also higher for the females than for the males, as shown by the larger value of standard error of the estimate and the lower value of R^2, the coefficient of determination of the corresponding regression model.

6.4.4 Ponderation on the adequacy of 10 as the numerical value for the rating of the maximal discomfort

There was an important issue to be solved before stating overtly the validity of the regression equation that expresses the overall relationship between the holding time and the discomfort ratings. It arose from the decision taken by the researcher to equate the maximal discomfort experienced by the subjects at the moment they stopped the holding with a rating of 10 rather than a higher one. Obviously, the end point of the scale has an important effect on the slope of the regression line, and with it on the interpretation of the subjective reactions to the exertion studied.

A straightforward test as to whether a value different from 10 could have been a better choice for the maximal discomfort rating was performed by removing from the dataset collected during each trial the discomfort rating obtained at the moment the subject decided to stop the effort, then fitting a regression line through this reduced dataset and comparing the slope of this line against that of the line fitted to the dataset that included the data point in question (the 'complete' dataset). This was in fact a comparison between the maximal value predicted by the actual data and the rating of 10 assumed by the researcher; if the slopes were significantly different, then the assumption made by the researcher should prove untenable. The values of slope of the regression lines fitted through the reduced sets of data are included in Appendix B, along with those of the lines already calculated for the complete sets.

Two-sample t-tests were used to probe for significance of the differences. Besides the comparison between the slopes of the 89 pairs of regression lines for the individual trials, comparisons were also made between the datasets divided in sub-samples according both to the abduction angle and to the gender of the subject. The results of the tests are shown in table 6.1.

Table 6.1 Results of the t-tests for the difference between the slopes of regression lines fitted to the data with maximal value of discomfort assigned as 10 (complete) and to the data without it (reduced). Data from 89 trials.

Samples being compared	Mean slope, complete dataset	Mean slope, reduced dataset	T-value	p-value
Whole	0.0988	0.1033	-0.86	0.39
Female subjects	0.1019	0.1047	-0.97	0.33
Male subjects	0.0956	0.0959	-0.15	0.88
Results at 30°	0.0986	0.0990	-0.39	0.71
Results at 60°	0.1000	0.1021	-0.66	0.51
Results at 90°	0.0977	0.0991	-0.42	0.68

Whichever way the sample was broken down, the slope of the regression line fitted to the reduced dataset was higher than that for the corresponding complete dataset. However, in every case the differences were far from significant, as shown by the p-values given in table 6.1. Furthermore, according to the regression model fitted using the reduced datasets, 10 is in fact the numerical value for the maximum discomfort that fits best with the discomfort ratings obtained up to the moment previous to the stoppage, considering that all the values predicted by the regression equations should be rounded up or down to 10.

This then demonstrated clearly that assigning the value of 10 to the degree of discomfort experienced by the subjects when they decided to stop the holding did not have a significant effect on the shape of the relationship between the holding time and the discomfort ratings. Therefore, the overall equation fitted to start from the origin is a valid expression of the way in which the subjects perceived their discomfort to grow with the passage of time.

The influence of the gender of the subject and of the abduction angle was also tested on the reduced datasets, and the results were similar to those already known for the complete datasets. The mean slope of the datasets collected from the trials on the female subjects was 0.1047, significantly higher than that for the males, 0.0959 ($t= 3.49$, $p=0.0008$). The mean slope for the datasets collected at each of the three abduction angles was 0.0997 at 30°, 0.1021 at 60° and 0.099 at 90°; these values did not differ significantly (ANOVA test, $F= 0.47$, $p= 0.626$).

6.4.5 Discomfort ratings at the upper end of the scale

The analysis of the discomfort ratings gathered during the second experimental stage raised an important issue concerning the values returned by the subjects as they approached their endurance limit. It has to be said again that the researcher did as much as he could to get the subjects familiar with the scale, emphasising the need to understand the meaning of the verbal expressions used to anchor the key values that appear on it.

However, an inspection of figures 6.1 (a) - (j) will show that, with very few exceptions, the subjects returned more than one rating of 10 before calling the effort to its end. This might appear somewhat undesirable since, according to Borg himself (Borg, 1990), the rating of 10 should be used to characterise the perceptual intensity elicited by a stimulus that the subject would identify as the strongest they have ever experienced. Knowing this, the researcher placed special emphasis on instructing the subjects to be conservative in their ratings so that they would not run out of scale on which to express their sensations while approaching their endurance limit. Therefore, when the early trials produced strings of discomfort ratings of 10 (in some cases preceded by a series of ratings of 9), the experimenter insisted to the subjects that they should avoid rushing into the high ratings; nevertheless, as more trials were completed, the phenomenon kept appearing. However, it has been shown that the data obtained from the 89 trials followed a very similar linear pattern, so that when treated either as a single sample or as a set of 89 separate samples, yielded correlation coefficients that were not only significant (with a single exception), but in most of the cases they were quite similar values. Therefore, if there was a bias in the way the subjects used the scale for the rating of discomfort, its nature was such that it did not affect the overall linear pattern of growth with the passage of time. This indicates that such bias, even if it existed, was the same for all the subjects and, rather than the evidence of some form of experimental error, it could constitute a feature of the way in which fatigue developed.

However, the repeated ratings at the upper end of the scale could be linked to a second methodological issue, which is not related to the rating scale per se; it might rather be the result of a biased perception by the subjects of what was expected from them. This bias might have occurred in either of two forms, or even both of them. On the one hand, it could be that the subjects tried to please the experimenter by 'going all the way', even if this meant enduring more discomfort than they actually should. On the other hand, their motivation could have been more mundane, just a desire to improve on their own past performance, or to compete against their fellow subjects. In fact, an excess of self-competitiveness was at play in the two trials that the experimenter himself had to stop, as was described in section 6.3.1. Nevertheless, this possibility had been acknowledged beforehand: from the moment they were being briefed about the aims of the investigation, even before they agreed to co-operate with it, it was made clear to the subjects that it was by no means a contest of any sorts, and they should not come to the laboratory thinking of lasting longer than someone else did, or longer than they themselves did the last time around. To reinforce this impression, the subjects were never told how long their holding time had been, even though some of them were quite insistent in trying to find out.

It is impossible to say whether or not the subjects were under such kind of motivation. However, the consistency of the holding times during the replication of the postures seemed to indicate that it was not so; true, on average, the maximum holding times at a given angle increased from trial to trial, but this was not so for every subject at the three angles. Besides, the subjects were repeatedly made aware that the aim of the research was not to see for how long they could stand the worst possible discomfort, but to find out how long it takes to reach that point. By stopping at roughly the same point during the repetitions of each posture, the subjects showed that they were well capable of recognising their endurance limit.

6.5 Conclusions

These may be put quite briefly as follows:

1) In the instance of purely postural exertion studied here, the discomfort grew in a linear fashion; discomfort ratings and holding times exhibited linear correlation coefficients as high as 0.996;

2) The strength of the linear relationship between discomfort ratings and holding time was not affected by the change in the abduction angle;

3) Although still strongly linear, the pattern of discomfort growth exhibited significant differences when compared between male and female subjects.

CHAPTER 7

ASSESSMENT OF MUSCULAR FATIGUE MANIFESTED BY THE CHANGES IN THE ELECTROMYOGRAPHIC SIGNAL

7.1 Introduction

The collection and analysis of electromyographic signals during the main experiment served four purposes:

i) to establish whether the holding of the posture until the appearance of unbearable discomfort provoked changes in the characteristics of the EMG signal which indicated the development of fatigue (first part of the fifth specific objective, chapter 3);

ii) if EMG changes were demonstrated, to assess the influence that the experimental conditions might have on their nature and extent (second part of the fifth specific objective);

iii) to investigate the nature of the relationship between the electromyographic indicators of the development of fatigue and the subjective perception of the increase of discomfort (third part of the fifth specific objective);

iv) still assuming the existence of significant EMG changes, establish whether these will persist beyond a certain time limit following the end of the holding trial (sixth specific objective).

This chapter reports on the procedures followed to accomplish those four goals. The work proceeded through the following stages:

a) collection of EMG signals before, during and after the holding;

b) spectral analysis of the signals to obtain values of mean power frequency (MPF) and RMS amplitude;

c) analysis of the change in MPF and RMS amplitude over the holding time;

d) study of the relationship between the time-related changes of MPF and RMS amplitude and the increase of discomfort ratings;

e) comparison of the reference EMG signals collected before the holding and after rest.

7.2 Procedures

7.2.1 Selection of muscles

Electromyographic signals were collected from the descending portion of the trapezius muscle and from the medial and posterior portions of deltoid muscle on both arms. The choice of these muscles was based on the reports of the location of maximum discomfort created by arm abduction, obtained from the eight female subjects who took part in the first experimental series.

The subjects always identified their right arm as the most uncomfortable, but this does not mean necessarily that the left arm was getting less fatigued. All the subjects were right-handed and this could make them more aware of their sensations on that side of the body. Therefore, it was decided that EMG signals would be collected from both arms.

7.2.2 Collection of EMG signals

Full details of the electromyographic equipment, the procedures followed for the location of the picking-up sites, the application of the electrodes, and the picking-up, conditioning and storing of the EMG signals have been given in chapter 3 and need not be repeated here.

During the trials, the first EMG signal was always collected between 5 and 10 seconds after the subject had adopted the required posture. This gap was allowed in order to reduce the possibility of picking up any surplus myoelectrical activity created by the act of bringing the arms up to the required position. But it had to be kept relatively short to ensure that no change attributable to fatigue would be missed, which could have easily occurred during the trials by the subjects with fairly short holding times, particularly when their arms were abducted at 60° or 90°.

After this initial recording, the signals were collected at intervals of 60 seconds, or 120 seconds at some stages in the course of the longer trials. The decision to allow this longer interval between consecutive recordings was taken after the analysis of the information collected during the earliest trials showed that during the longer trials neither the EMG signal nor the discomfort ratings being collected at the same time presented significant modifications between consecutive recordings separated by 60 seconds. However, even in those trials signals were collected every 60 seconds for at least the first 10 minutes into the holding and in the period when the experimenter judged, based on the discomfort ratings, that the subject was getting close to the moment when the discomfort would become unbearable. In every case, the final sample was collected precisely at the moment when the subjects reported to have reached that subjective limit. The sampling rate was 1112 Hz, each sample was 2.23 seconds long and contained 2480 data points.

7.2.3 Spectral analysis

As mentioned in the review of the literature, muscular fatigue is generally accepted to be shown by either of two changes in the EMG signal: a shift of the frequency components towards lower values, or an increase in the value of the RMS amplitude of the signal (De Luca, 1985). Both criteria were tested in this investigation, by looking at the values of mean power frequency (MPF) and RMS amplitude for every sample collected during the holding of the postures. The value of MPF was calculated from the spectral power density function derived by Fast Fourier Transformation of a section of the sample that contained 2048 data points, which were included between 0.19 and 2.04 seconds of the sampling period. The value of RMS amplitude was calculated from the section collected between 0.5 and 2.0 seconds of the length of the sample, containing 1670 data points. Leaving out a number of data points at both ends of the sample reduces the chance of feeding into the calculation any form of artifact created at the start and the end of the recording process. The spectral analysis was performed using the DaDISP software package (DSP Development Corp., Cambridge MA, USA).

7.2.4 Reference muscular contractions

Assuming that the holding of the posture did in fact provoke changes in the EMG signal which indicate the presence of muscular fatigue, it is obviously important to try and establish for how long after the cessation of the effort those changes will remain. A simple and straightforward means of getting that information is to compare the characteristics of the EMG signals collected when the muscle is required to exert, before and after the postural effort, a force that should activate it to the same extent. To that end, the subjects were asked to perform a series of manoeuvres designed to activate, one at a time, each of the muscles whose response to arm abduction would be evaluated. These manoeuvres, called reference muscular contractions (RMC), were performed by holding a known weight in a manner that elicited an effort directly on the muscle being tested.

The movements used to activate the muscles under study were carried out following the procedures suggested by Janda (1983). To activate the trapezius muscle, the subjects held the weight in their hand with the arm fully extended by their side. Whilst the researcher placed a hand on their shoulder to fix the clavicle, the subjects were instructed to pull upwards (simulating a shrugging action) against the resistance of the weight. To activate the posterior deltoid, the subjects held the weight with their arm abducted at 90° in the coronal plane and the elbow flexed also at 90°, placing the forearm parallel to the ground; in this position, they were instructed to push backwards against the researcher's hand placed at the back of their upper arm, just above the elbow. Finally, the medial deltoid was activated by the sole action of the weight being held by the subject with their arm abducted 90° in the coronal plane and in full extension. During the trials, in order to allow a full EMG sampling period, the subject had to sustain the reference contraction for a minimum of 3 seconds. Figures 7.1 a), b) and c) show one of the male subjects performing the RMC for trapezius, medial and posterior deltoid, respectively.

The weights used to obtain the reference muscular contractions were adjusted to each muscle of each subject, and are listed in table 7.1. To find those weights, during their first visit to the laboratory the subjects were asked to perform the manoeuvres described above, using a device whose total weight could be varied between a minimum of 2 kg and a maximum of 11 kg. The subject started by holding the minimum of 2 kg and this was increased by 0.5 kg at a time, until reaching a weight they could not sustain for the 3 seconds required; that weight minus 0.5 kg was the one they used. The weight-holding device may be seen in the three illustrations presented in figure 7.1.

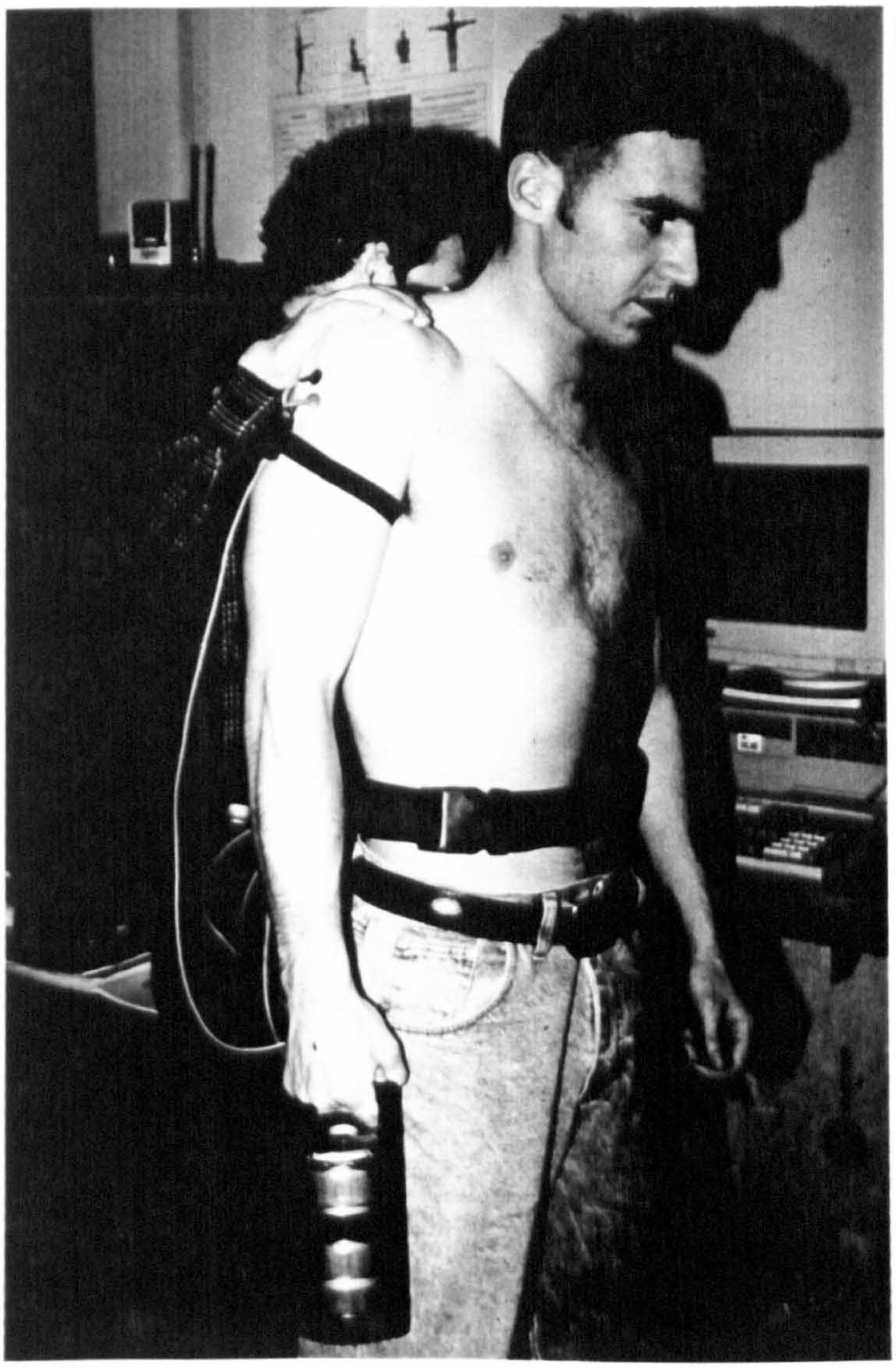


Figure 7.1 a) A subject carrying out the manoeuvre devised to obtain the reference muscular contraction of the trapezius muscle.

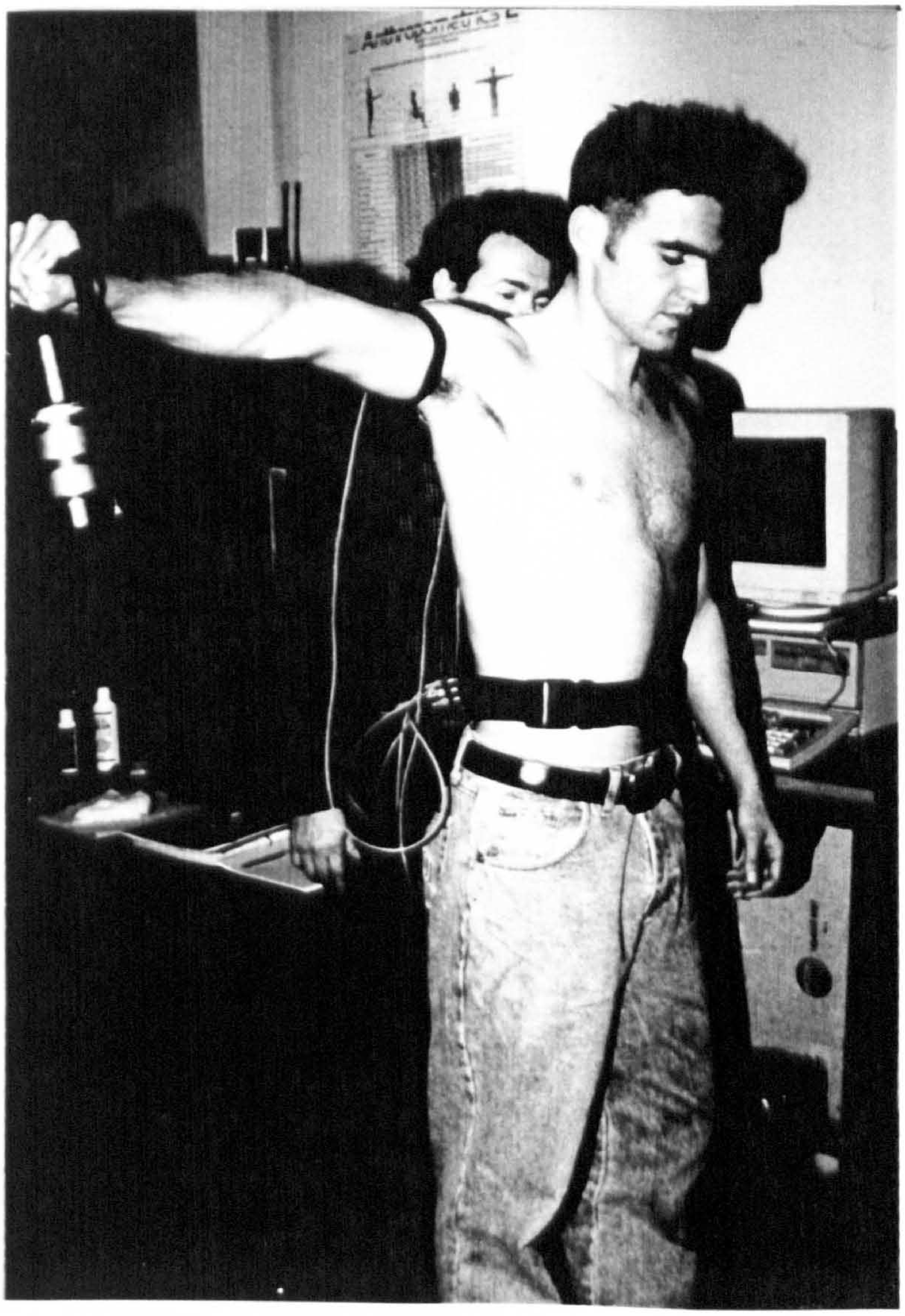


Figure 7.1 b) A subject carrying out the manoeuvre devised to obtain the reference muscular contraction of the medial deltoid muscle.

Figure 7.1 c) A subject carrying out the manoeuvre devised to obtain the reference muscular contraction of the posterior deltoid muscle.

Table 7.1 Weights (kg) used by each subject to perform the reference muscular contractions of each muscle.

	Right Shoulder			Left Shoulder		
Subject	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
1	11.0	3.0	3.0	11.0	3.5	3.5
2	11.0	3.5	3.5	11.0	3.0	3.0
3	11.0	3.5	3.5	11.0	3.0	4.5
4	8.0	2.5	3.5	8.0	3.0	4.0
5	8.0	3.0	4.5	9.0	3.0	4.5
6	11.0	4.5	5.5	11.0	4.5	5.5
7	11.0	7.0	8.0	11.0	7.0	8.0
8	11.0	4.5	5.5	11.0	4.0	7.0
9	11.0	5.0	6.5	11.0	4.5	6.5
10	11.0	5.0	6.5	11.0	5.0	6.5

In each trial, once the electrodes were in place, the subjects performed an RMC for each muscle, rested for 10 minutes, and then held the required posture to their limit of endurance. After they halted the exertion, the subjects rested for 5 minutes, and then performed a second RMC.

Although there was no way of checking whether it was long enough to permit the subject a full recovery, the duration of 10 minutes assigned to the rest period following the completion of the RMC manoeuvres prior to the holding was in line with the usage in studies similar to the present one, for example, Viitasalo and Komi (1977), Gerdle et al (1988), Daanen et al (1990), Caffier et al (1993). Furthermore, since all the cited authors have asked their subjects to carry out the maximal voluntary contraction (MVC), which involves a larger force than the one used by the subjects in this study, it is most likely that the rest was sufficient to give the subject opportunity to recover. The length of the rest following the cessation of the effort was decided in very much the same fashion, since it has been reported that the characteristics of the EMG signals returned to their pre-effort values within a period of 5 minutes following exertion (Petrofsky and Lind, 1980; Mills, 1982; Merletti et al, 1983; Kuorinka, 1988).

7.3 Results

7.3.1 Main features in the change of the EMG activity over the holding time

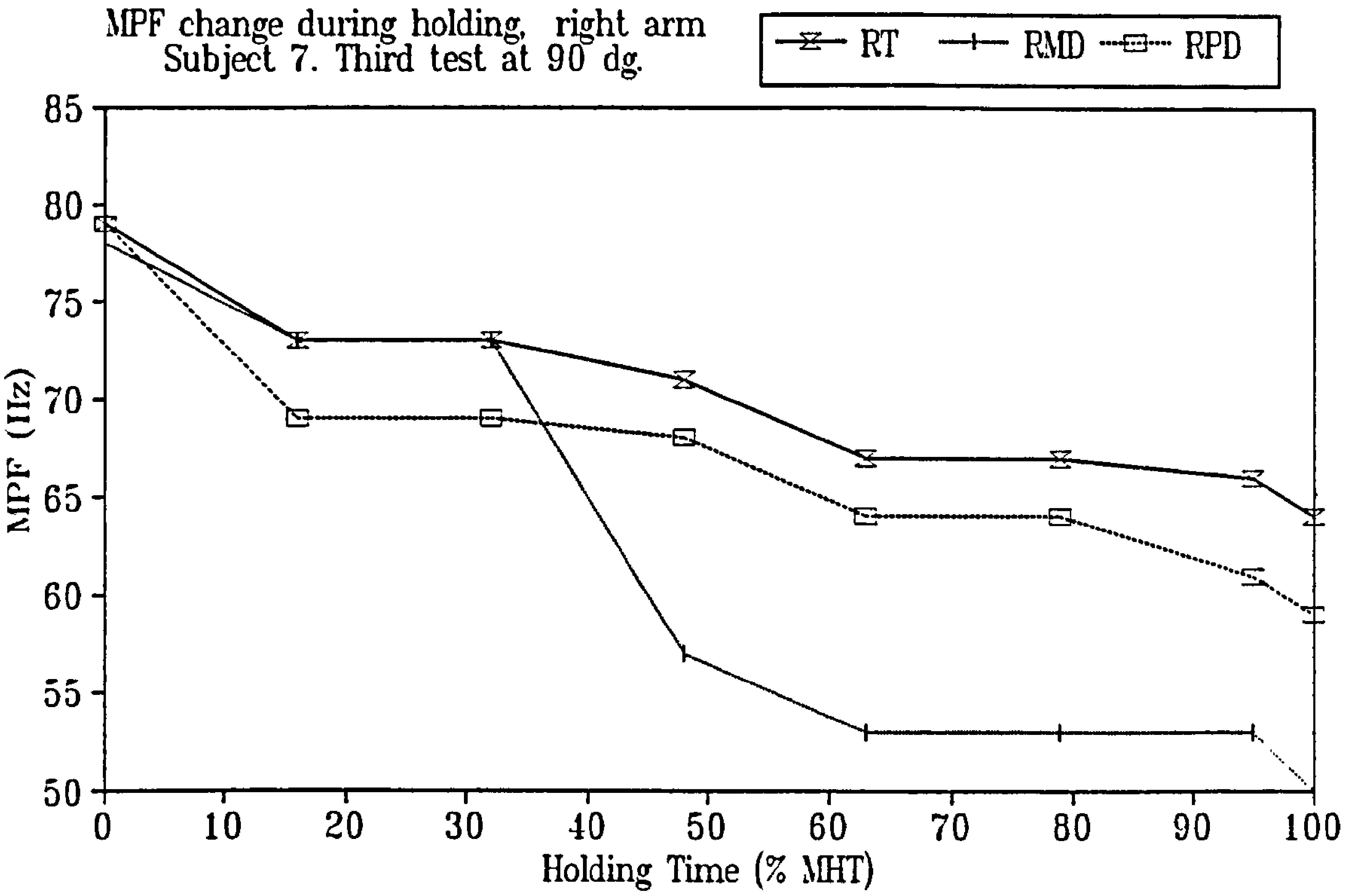
As a first approach to the study of the changes occurring in the muscles during the holding of the postures, the values of MPF and RMS amplitude calculated from the EMG signals collected during each of the 89 trials were plotted against the corresponding sampling times.

The plots of the EMG changes generated during the nine trials performed by each individual (only eight trials for subject No. 6) were put together and studied visually. This inspection showed that, even though there were important differences in the way the muscles responded to the three abduction angles studied, each individual responded in a very similar way during the repeated trials at a given angle. Thus, if the plots prepared for all 89 trials were included in this thesis, they would not necessarily amount to a better description of the changes in the EMG signal that were provoked by the holding of the postures. Therefore, only the graphs depicting the EMG changes that occurred in the course of one of the trials completed by each subject at each abduction angle (a total of 30 trials) are presented; each one of those graphs is an adequate representation of what happened during the other two trials at that particular angle. These graphs may be found in Appendix D . The representative trials chosen are those with the largest coefficient for the correlation between the discomfort ratings and the holding time (which were calculated in chapter 6 and are included in Appendix B). In the graphs presented in Appendix D , the changes in MPF have been plotted separately from the changes in RMS amplitude, and separate plots are also presented for the muscles of each arm. Typical examples of plots are used in this chapter to illustrate the salient features in the EMG responses.

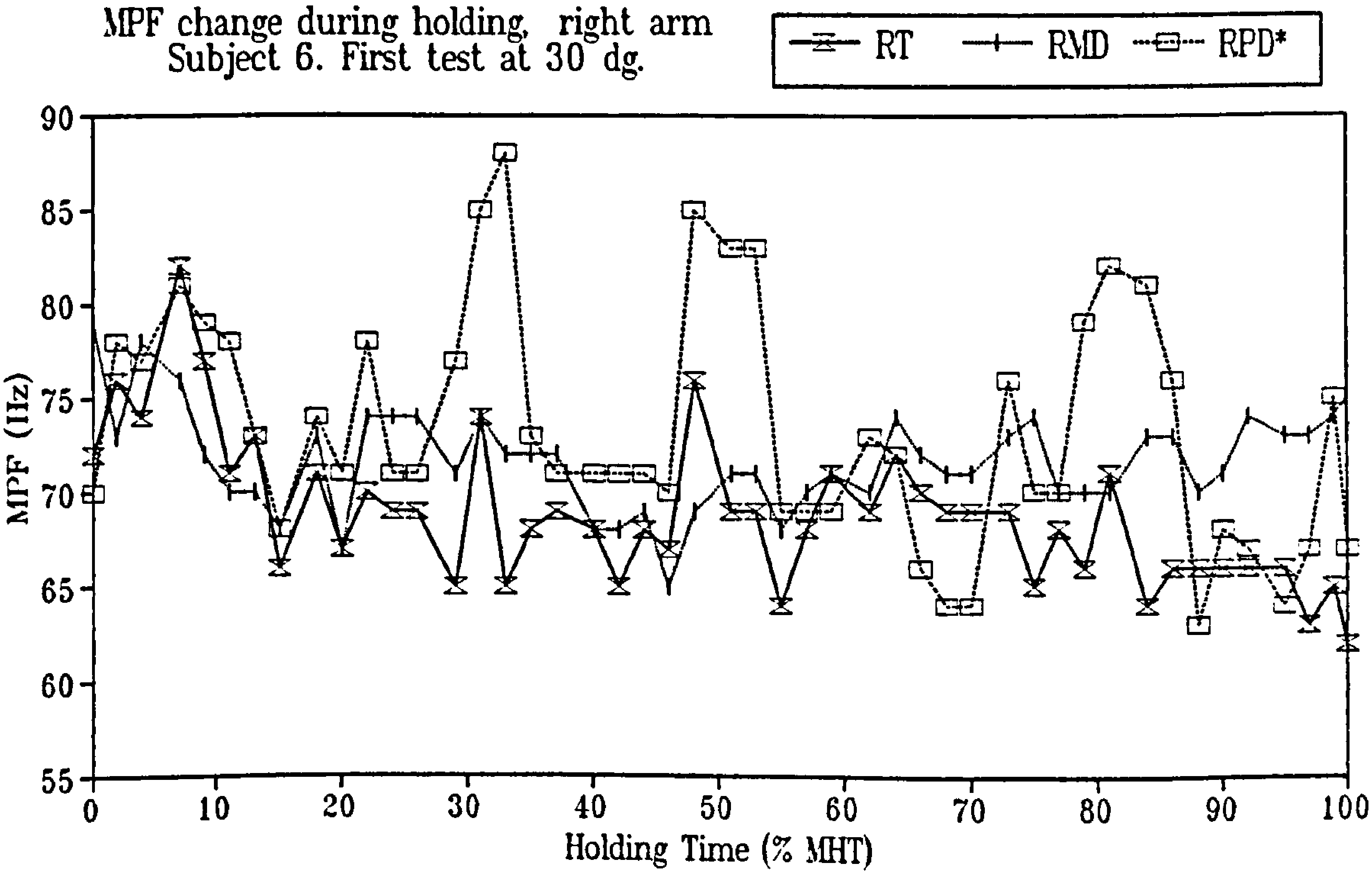
In order to establish a common temporal basis on which to study the features of the EMG responses, the holding times were normalised and expressed as percentage of the maximum holding time (% MHT). As already shown in chapter 6 when dealing with the discomfort ratings, this normalisation of the sampling times reduces the influence of the variations of the actual holding times on the other variables studied.

A look at the plots of the values of MPF and RMS amplitude against the holding time included in Appendix D will show that in most of the trials the expected fatigue-related changes did occur, in that as the holding progressed, the values of MPF tended to decrease and those of relative RMS amplitude tended to increase. However, only in a few cases those changes exhibited a fairly smooth, well-defined pattern, which one could expect to be of a reasonably similar shape for the three muscles. Figure 7.2 a) and c) show samples of the changes that exhibited this 'exemplary behaviour'. In quite a sharp contrast, in the majority of cases (and particularly for MPF) the changes appeared to happen almost at random, the values going up and down from one measurement to the next with wide differences between the muscles. Figure 7.2 b) and d) illustrate this kind of behaviour. The pattern of fatigue-related changes was not therefore as clear-cut as had been expected.

The graphs included in figure 7.2 a)- d) also serve to illustrate the fact that a wide variation in the pattern of change of the EMG signal was more evident in the course of trials where a long MHT was achieved. For all the subjects the 'wild' pattern appeared more clearly at an abduction of 30° than at 60° or 90°, and comparing subjects at the same abduction angle, the 'wildest' of those patterns were evident for subjects numbers 10, 9, 6 and 3 who, in that order, had on average the longest holding times.

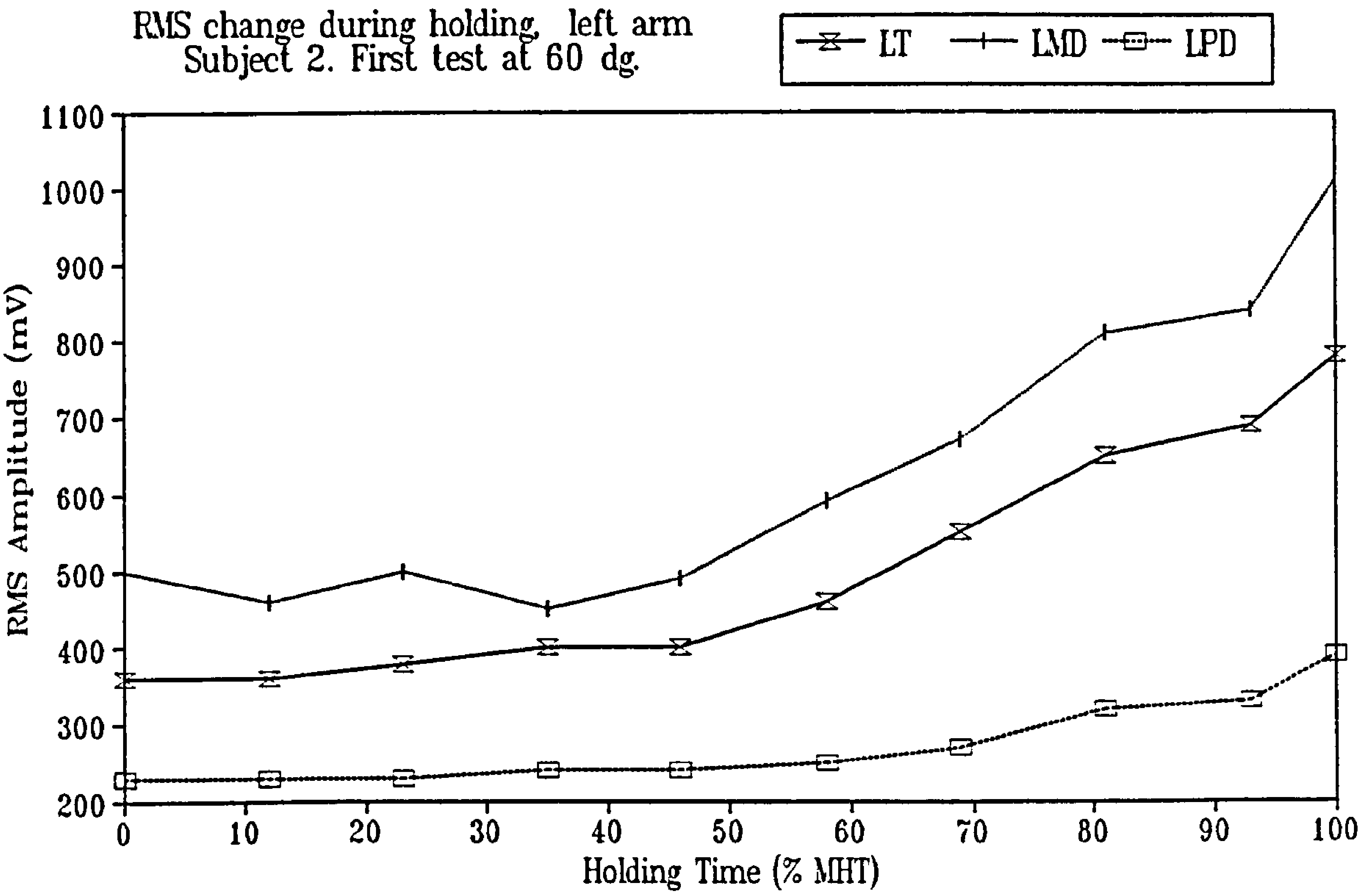


a) Similar, smooth pattern of change in MPF for right trapezius and right posterior deltoid

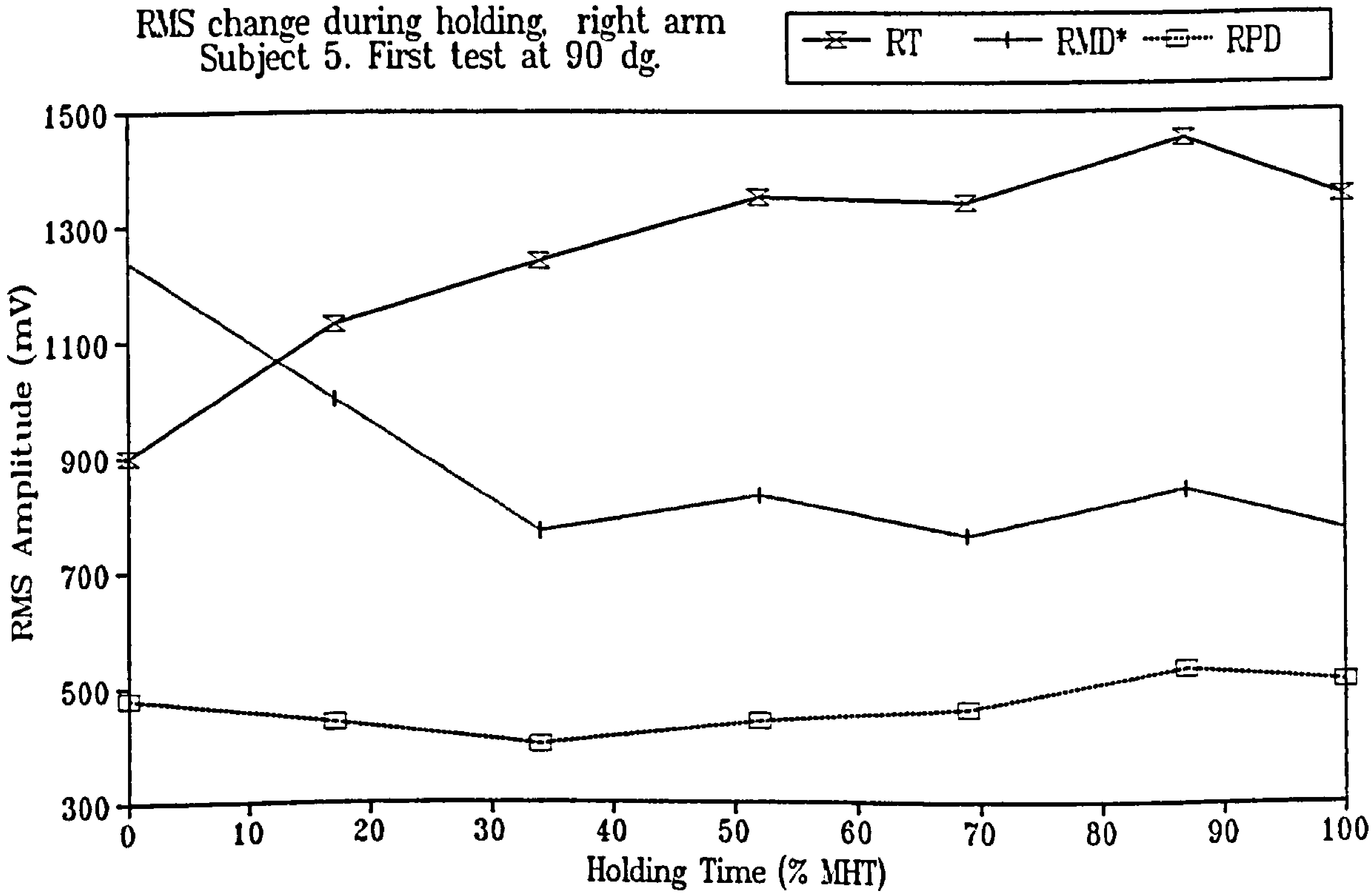


b) MPF Changes with a different, unsmooth pattern for the three muscles of the right arm

Figure 7.2 Examples of the variability of the pattern of EMG changes over holding time



c) Changes in RMS amplitude with a similar, smooth pattern for the three left muscles.



d) RMS amplitude changes with a different, unsmooth pattern for the three right muscles.

Figure 7.2 Examples of the variability of the pattern of EMG changes over holding time.

7.3.2 Presence of reversed changes in the EMG parameters

Another fact evidenced by the plotting of the values of MPF and RMS amplitude against the holding time is the existence of a number of cases where the changes went in direction opposite to that expected: that is, MPF tended to increase instead of to decrease, and RMS amplitude behaved in exactly the opposite fashion. The number of cases in which such reversed changes occurred in each of the six muscles studied is shown in table 7.2.

Table 7.2 Number of trials, out of a total of 89, in which the features of the EMG signal collected from each of the muscles studied changed in direction opposite to that expected.

	Right Shoulder			Left Shoulder		
Reversed change	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
MPF increase	31	4	2	24	1	4
RMS amplitude decrease	0	23	2	2	22	8
Both	31	27	4	26	23	12

The presence of the reversed changes has been taken up at this point only for the purpose of presenting a full description of the electromyographic phenomena observed in the course of this experiment. The remarkable features exhibited by those reversed changes, and their obvious links with the factors that made up the experimental design will be considered in the Discussion section of this chapter. In addition, the possible factors behind their existence, and the implications of this rather unexpected occurrence will be reviewed at length in the Discussion chapter.

7.3.3 Level of activation of the muscles during posture holding

Whilst the changes in RMS amplitude indicate the presence of muscular fatigue, the actual value of this parameter provides an indication of how activated is the muscle during the isometric contraction. This is so because, in this mode of exertion, the value of RMS amplitude is directly related to the number of muscle units recruited at the moment the signal was recorded (De Luca and Knaflitz, 1992). Therefore, by comparing the time course of the value of RMS amplitude of the three muscles studied on each arm it is possible to determine which of them was the most heavily activated throughout the trials of posture holding.

In the situation under study, however, a comparison of the actual values of RMS amplitude would be practically useless, since all that might be inferred from it is what muscle was recruiting the most units, but this would not tell what proportion of its strength was being used. In order to determine this, the value of RMS amplitude calculated from the signals recorded throughout the exertion was normalised against the RMS amplitude obtained during the reference muscular contraction (RMC) performed prior to the holding, which amounted to the selective activation of each muscle to a level that (ideally) should be the same every time the manoeuvre was performed.

Thus, the RMS amplitude extracted from each of the 1495 signals recorded during the 89 trials of maximum holding time was normalised against the RMS amplitude of the corresponding RMC, and converted to a percentage of this reference value. The mean, standard deviation and range of the percentage level of activation for each muscle is shown in table 7.3.

Table 7.3 Statistical features of the percentage level of activation exhibited by the muscles under study throughout the 89 trials for measurement of MHT. The individual values were calculated as percentage of the reference activation performed prior to the corresponding trial.

	Right Shoulder			Left Shoulder		
Statistical feature	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
Mean value	61.4	34.2	29.6	52.6	34.3	21.0
Standard deviation	22.42	12.26	19.97	21.96	12.14	14.44
Range	21 - 132	12 - 92	3 - 100	10 - 177	11 - 97	7 - 86

This table shows that in both arms it was the trapezius muscle the one which, on average, reached the highest levels of activation relative to the reference value, its average levels of activation were at least one and a half times those exhibited by the medial deltoid, and doubled those observed in the posterior deltoid. Besides, the range of the actual values indicates that there were cases where the activation of both trapezius muscles went well beyond the reference bench mark.

Another noteworthy feature of the information presented in table 7.3 is that the medial deltoid muscle exhibited quite consistent levels of activation. On average, it reached practically the same level in both arms, with the values spread over nearly identical ranges and with a very similar dispersion across it.

Figure 7.2 d) serves well to illustrate the disparity in the activation of the three muscles: it presents a clear gap between the trace for the trapezius muscle and those for the other two; however, this relative position is not present in figure 7.2 c). More examples of both situations may be found in the representative plots included in Appendix D . These show that the instance where the trapezius muscle shot well above the other two was more frequent during trials at 60° and 90°.

7.4 Statistical analysis

It has now been shown then that the postural exertion did in fact provoke changes in the myoelectrical activity which are compatible with the existence of fatigue, and this fulfils the first of the four purposes for which the collection and analysis of the EMG signals was undertaken. It also has emerged that in many cases those changes were not as smooth and well-defined as was expected from them. Now, attention can be turned to the study of the underlying causes of fatigue development and the course it followed, especially to the role that the modifications operated in the experimental conditions could play in the effects observed. Doing this will lead to the completion of the second purpose served by the study of the EMG signals.

7.4.1 Fitting of a model to the development of fatigue

For a start, given that changes indicative of fatigue in MPF and RMS amplitude did occur consistently during all the trials of posture holding, an attempt was made to establish whether those changes could be represented by a linear model. A regression line was fitted to the data collected during each of the 89 trials and, as the features of the time course of the EMG changes (discussed in section 7.3.1) would lead to expect, the results from this procedure varied widely. For all the six muscles there were cases where the fit of the regression line was quite acceptable, with a highly significant slope coefficient and R^2 reaching values well above 90%. However, in many of the cases where the values of the EMG parameters varied widely, despite the slope coefficient being significantly different from zero (i.e. the F^* test for the regression model reached significance), the variations around the regression line tended to cancel each other, yielding values of R^2 close to zero. Table 7.4 presents, for each muscle, the number of trials in which the t-test on the slope coefficient of the regression line, as well as the mean, standard deviation (S.D.) and coefficient of variation (C.V.) of the coefficients of determination (R^2) for all 89 trials. To avoid confusion with the value of the coefficient of determination, C.V. is given as a decimal figure rather than the customary percentage figure.

Table 7.4 Number of trials, out of the 89 performed, in which a linear regression model adequately described the changes in the EMG signal. The main statistical features of the coefficient of determination (R^2), calculated over all the 89 trials, are also given.

	Right Shoulder			Left Shoulder		
EMG feature	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
MPF	68 trials	85 trials	76 trials	58 trials	79 trials	70 trials
Mean R^2	56.9%	78.8%	70.6%	48.3%	69.1%	62.9%
S.D.	29.91	18.89	26.46	32.75	27.61	27.9
C.V.	0.53	0.24	0.37	0.68	0.4	0.44
RMS amplitude	77 trials	63 trials	64 trials	83 trials	69 trials	73 trials
Mean R^2	76.4%	57.7%	58.6%	83.4%	61.1%	67.9%
S.D.	24.35	31.87	33.94	19.93	31.07	28.02
C.V.	0.32	0.55	0.58	0.24	0.51	0.41

The extent of the variability in the EMG responses is well evident in the information presented in table 7.4. Its most obvious manifestation comes precisely through the quite appreciable size of the coefficient of variation, which for all the muscles is much higher than the 10% considered acceptable. The wide variability is also evident in the fact that whilst for all the muscles the regression model reached significance in a fairly large proportion of trials, only for the values of MPF in the right medial deltoid and of RMS amplitude in both trapezius muscles did the mean value of R^2 indicate the presence of strong linearity in their pattern of change. It is also in the left trapezius muscle that a third effect is visible, since whilst its average R^2 for RMS amplitude was the largest of all and had the smallest standard deviation, for MPF it was the shortest and with the largest s.d. Further interesting contrasts in the behaviour of the two EMG parameters will be highlighted later in this chapter, when their suitability as indicators of the extent of muscular fatigue is taken up for discussion.

Extreme examples of the variability in the adequacy of the regression model to follow the pattern of change of the EMG parameters are illustrated in figure 7.3 a) - l). This shows, for each of the six muscles, the traces corresponding to the EMG changes that exhibited the best and the poorest fit of the linear regression model. The identifier of the trials where these examples came from has been placed next to the corresponding trace. The changes in MPF have been plotted separately from those in RMS amplitude, and the corresponding regression equation in a generic format ($Y = b_o + b_1x$) has been included.

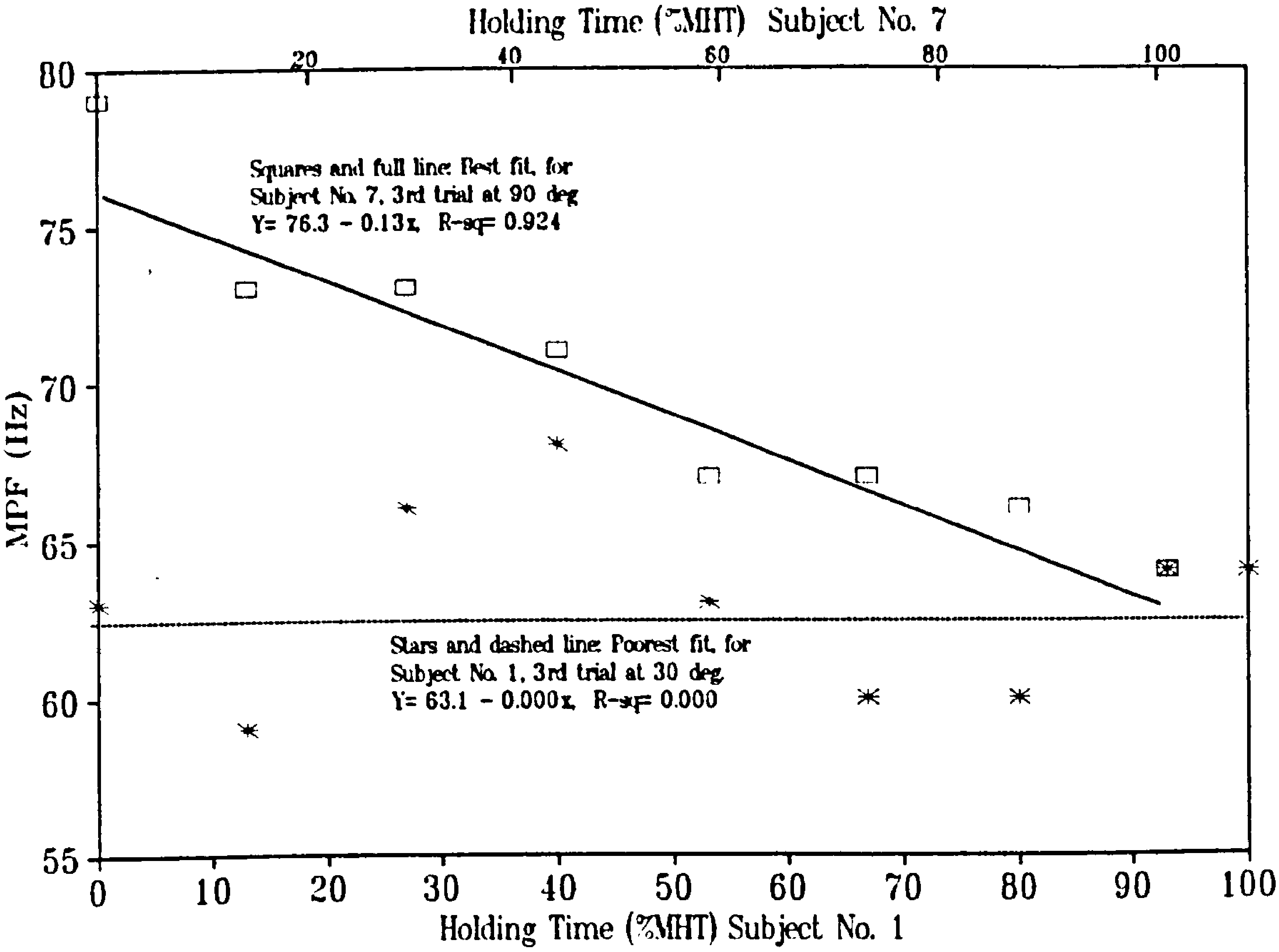


Fig. 7.3a) Extremes of the goodness of fit of a linear regression model to the change of Mean Power Frequency in the right trapezius muscle.

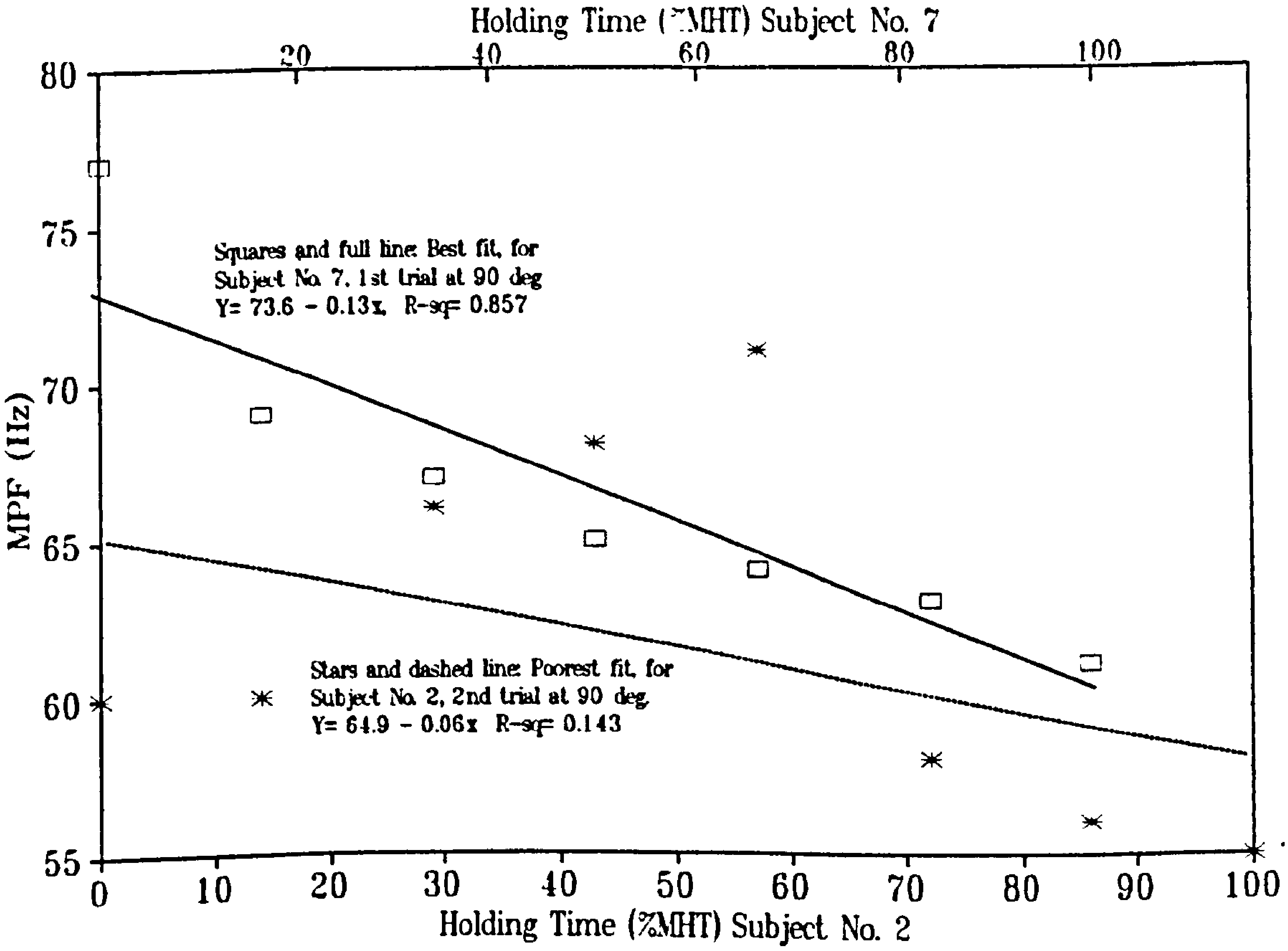


Fig 7.3b) Extremes of the goodness of fit of a linear regression model to the change of Mean Power Frequency in the left trapezius muscle

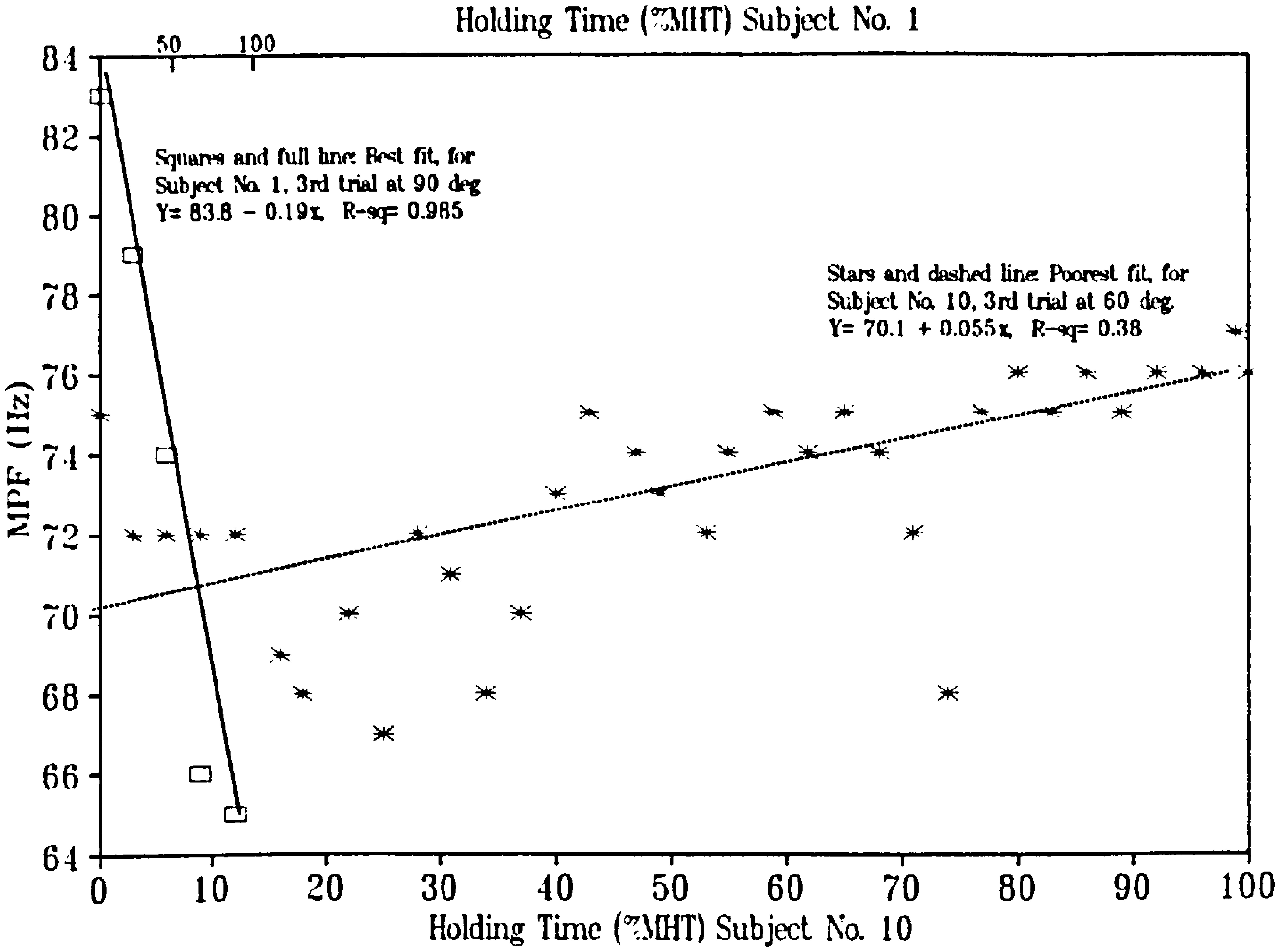


Fig 7.3c) Extremes of the goodness of fit of a linear regression model to the change of Mean Power Frequency in the right medial deltoid muscle

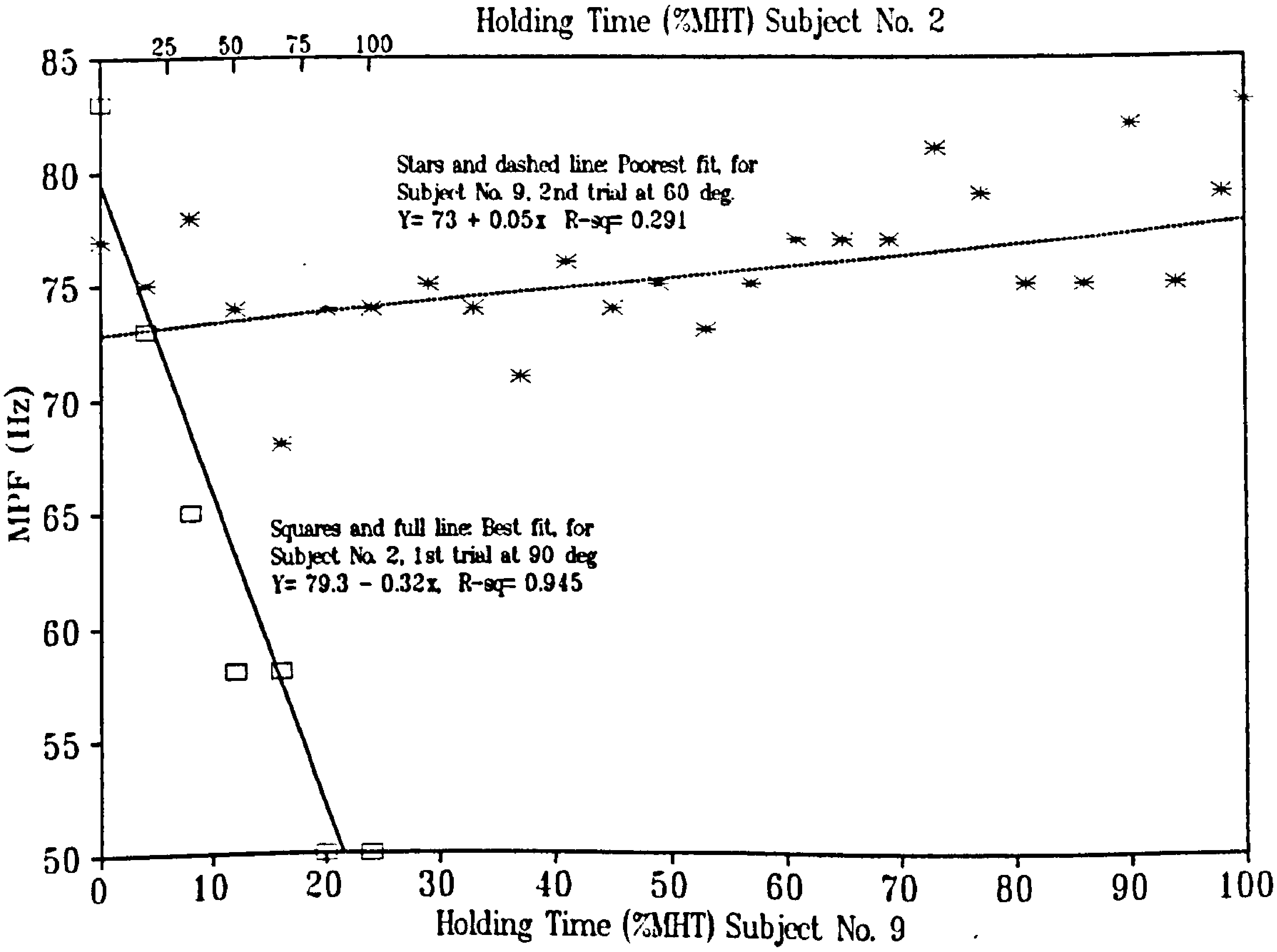


Fig 7.3d) Extremes of the goodness of fit of a linear regression model to the change of Mean Power Frequency in the left medial deltoid muscle.

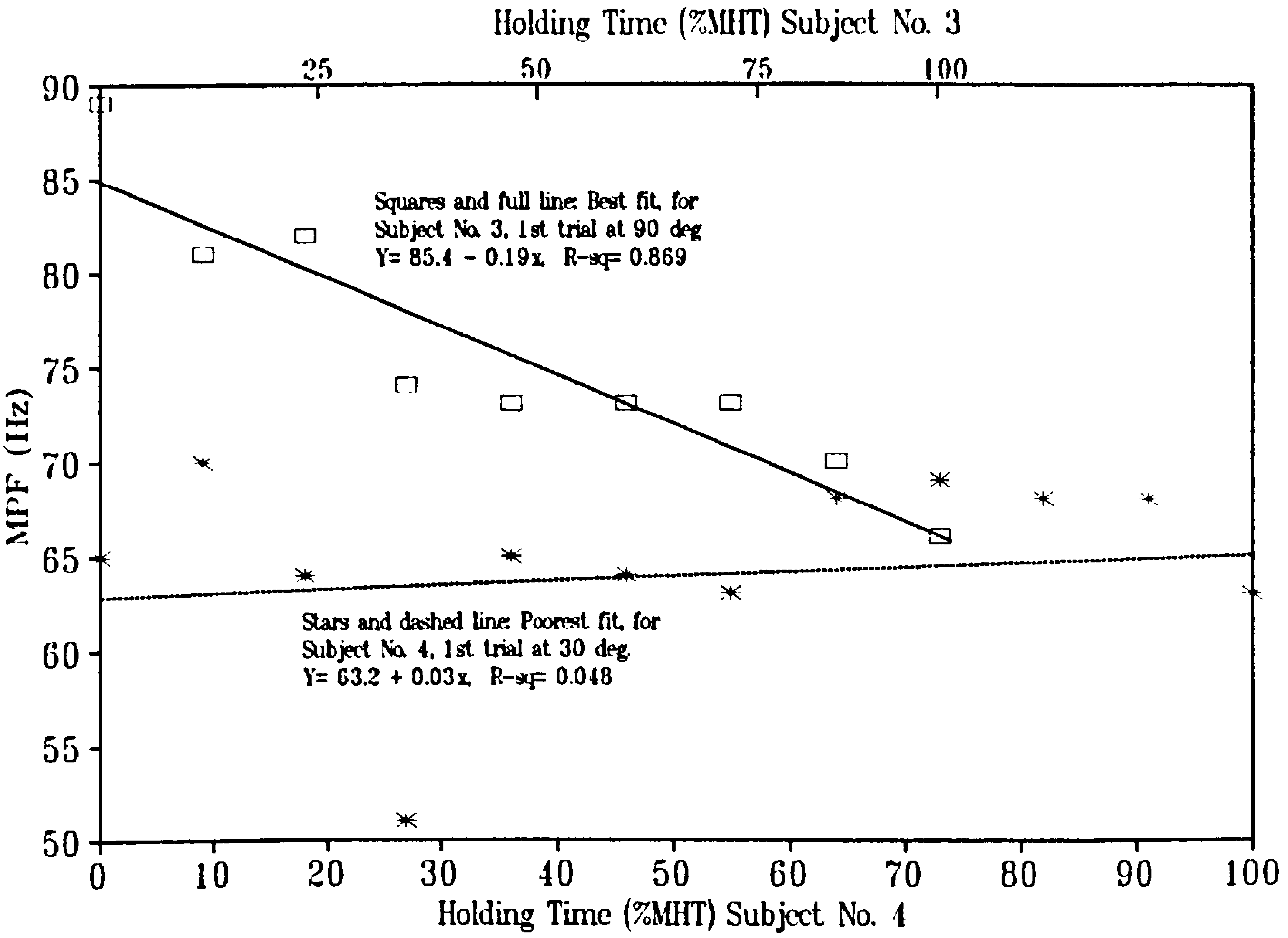


Fig 7.3e) Extremes of the goodness of fit of a linear regression model to the change of Mean Power Frequency in the right posterior deltoid muscle

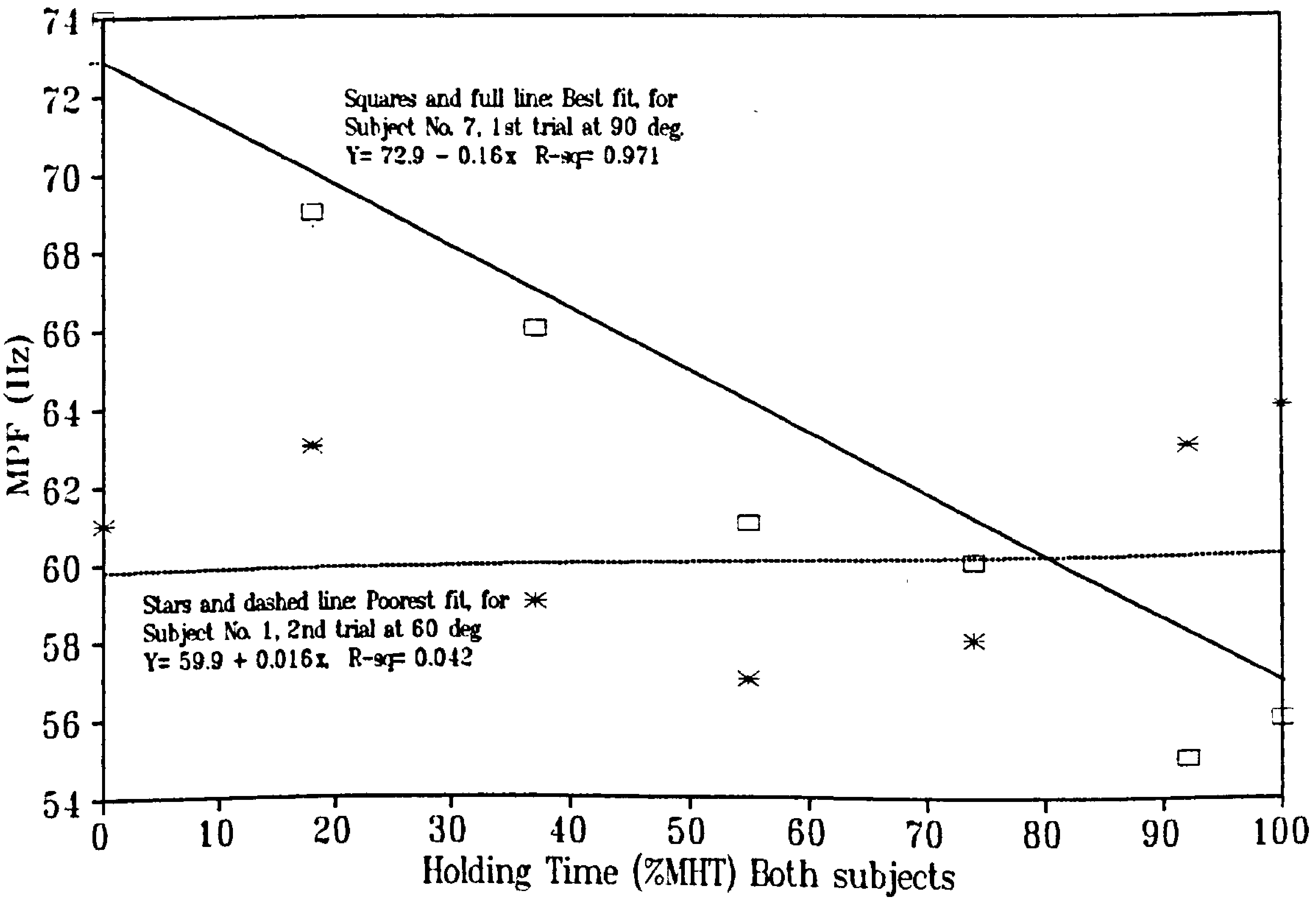


Fig 7.3f) Extremes of the goodness of fit of a linear regression model to the change of Mean Power Frequency in the left posterior deltoid muscle

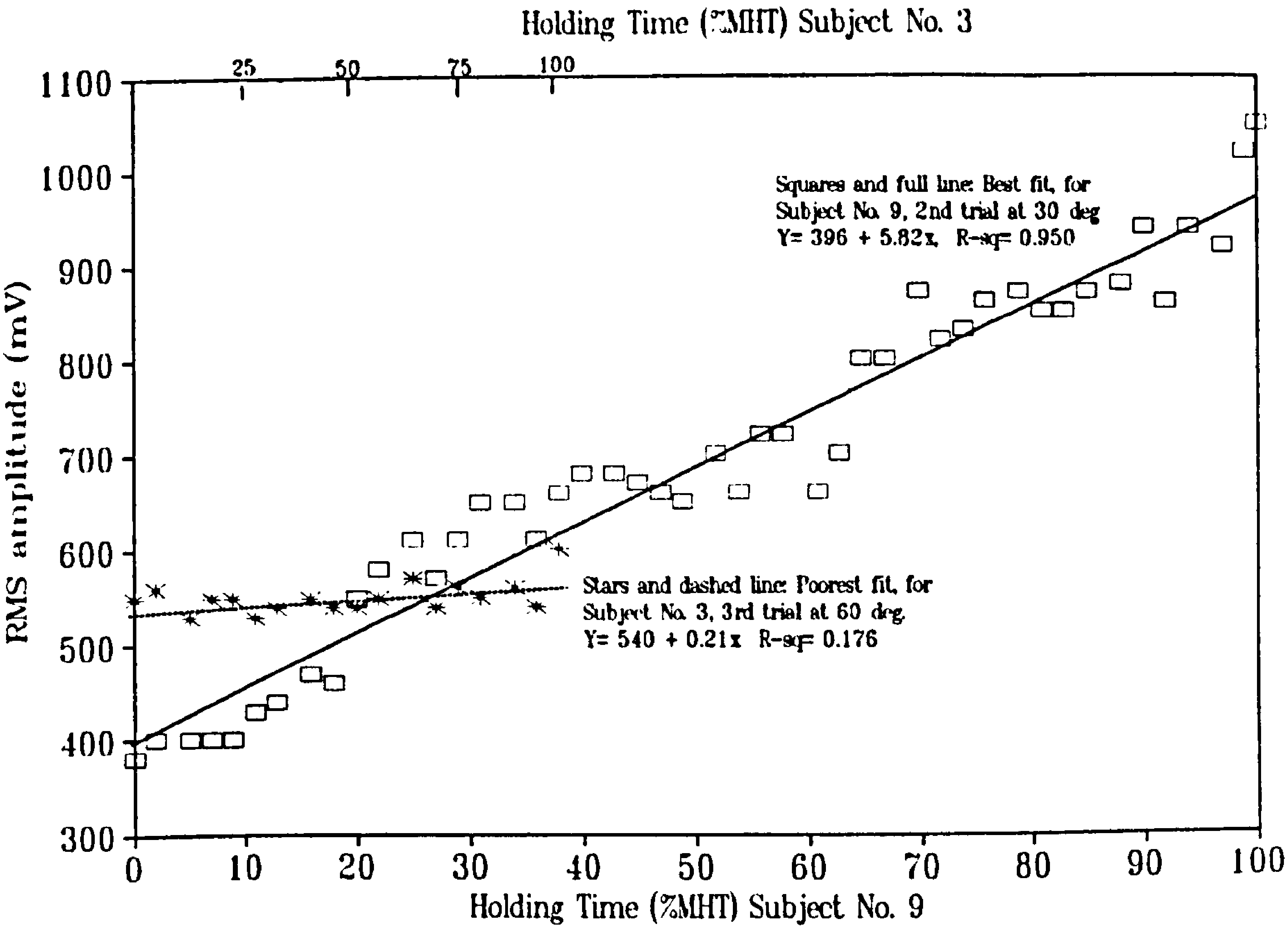


Fig 7.2g) Extremes of the goodness of fit of a linear regression model to the change of RMS amplitude in the right trapezius muscle

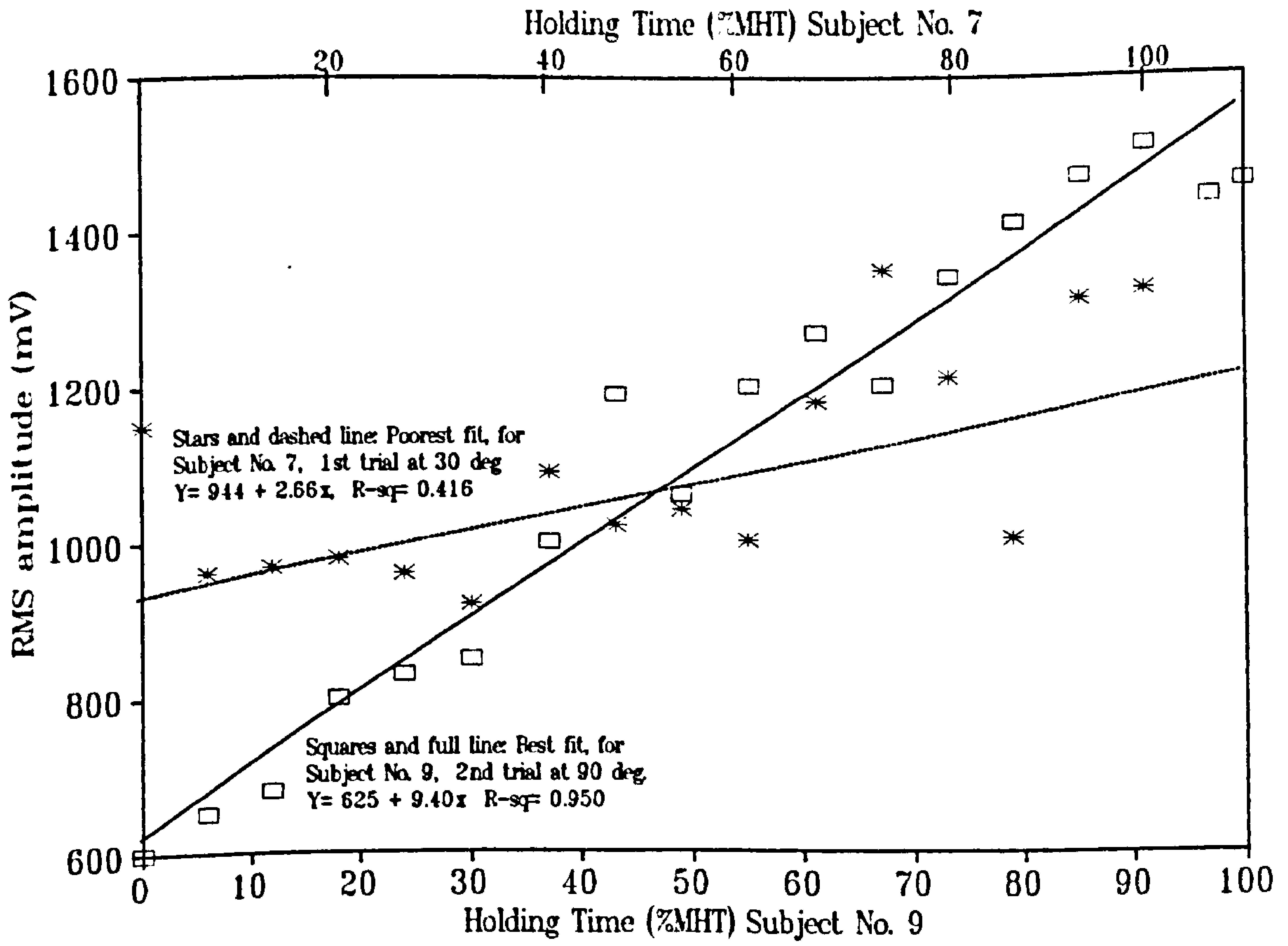


Fig 7.3h) Extremes of the goodness of fit of a linear regression model to the change of RMS amplitude in the left trapezius muscle

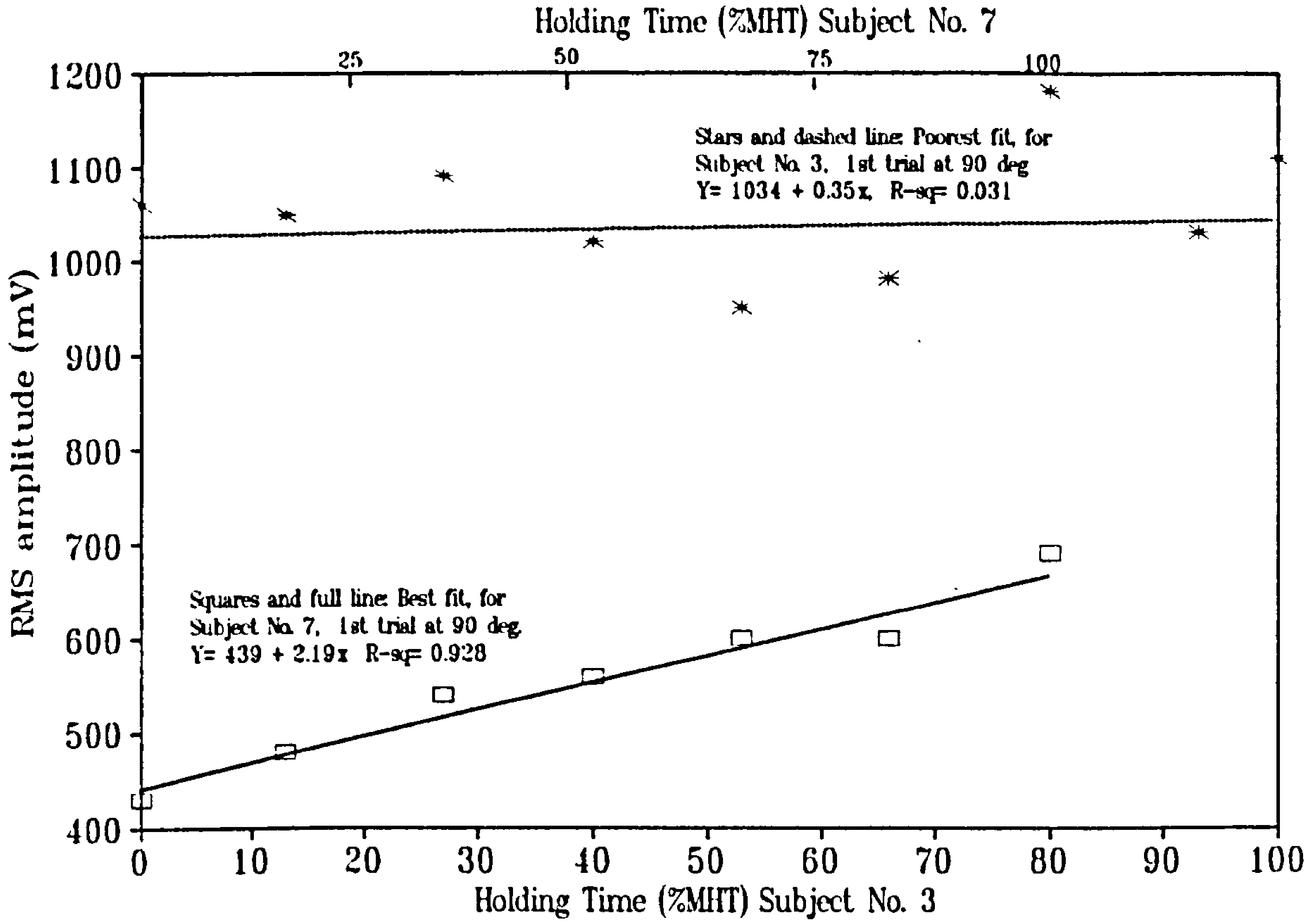


Fig 7.3i) Extremes of the goodness of fit of a linear regression model to the change of RMS amplitude in the right medial deltoid muscle

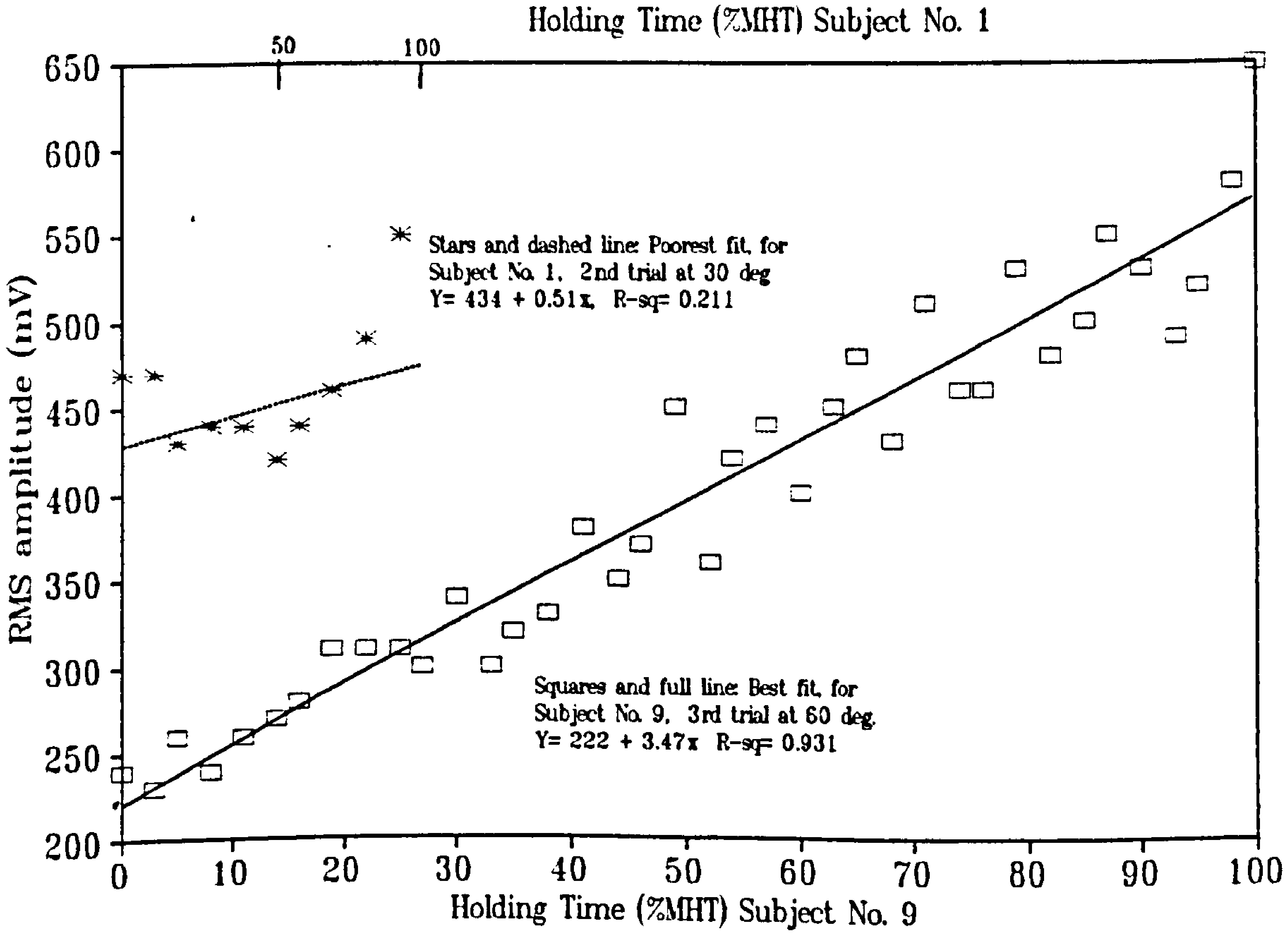


Fig 7.3j) Extremes of the goodness of fit of a linear regression model to the change of RMS amplitude in the left medial deltoid muscle

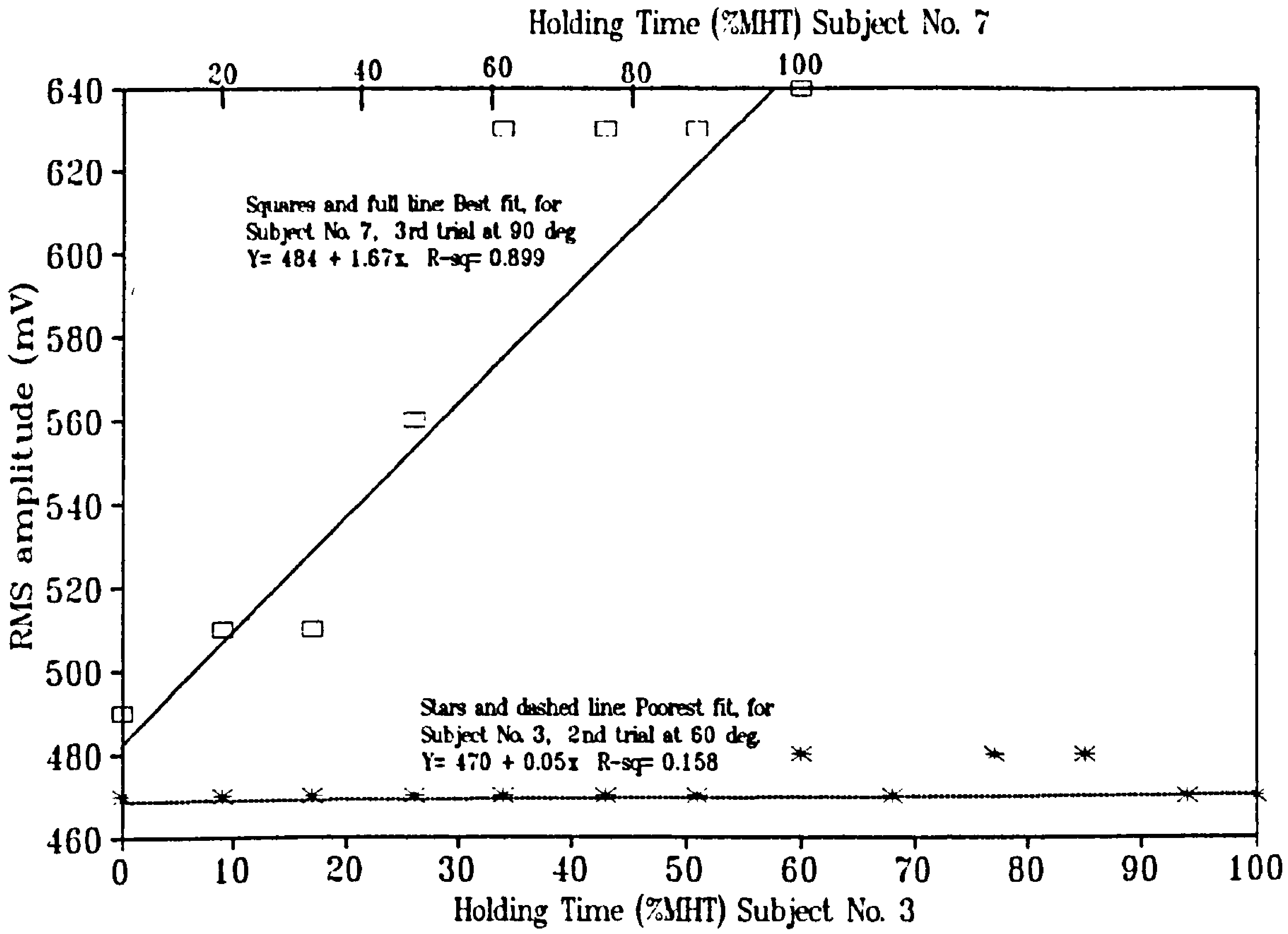


Fig 7.3k) Extremes of the goodness of fit of a linear regression model to the change of RMS amplitude in the right posterior deltoid muscle

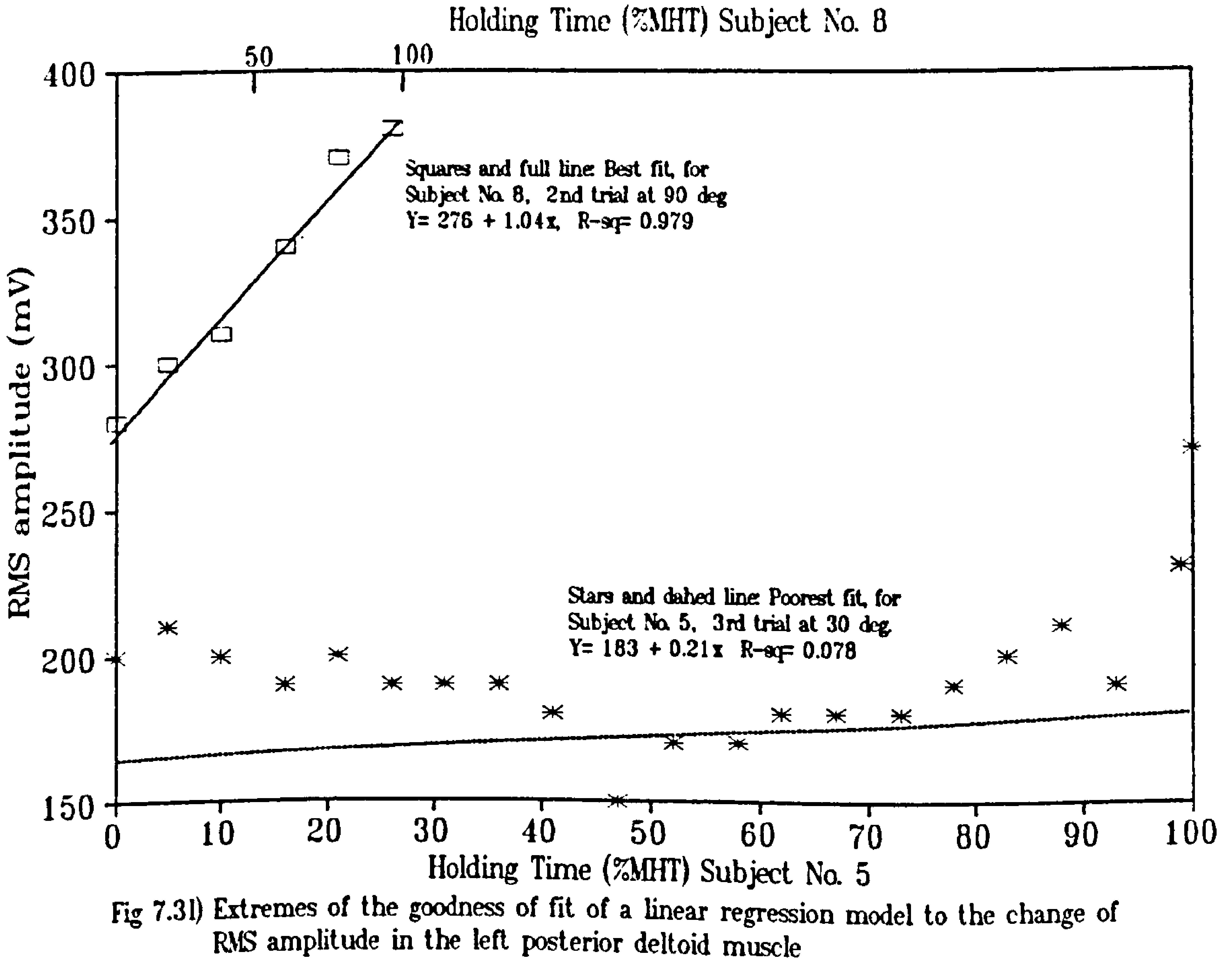


Fig 7.3l) Extremes of the goodness of fit of a linear regression model to the change of RMS amplitude in the left posterior deltoid muscle

7.4.2 Influence of gender and body side on the starting values of MPF and RMS amplitude.

The visual inspection of the representative plots included in Appendix D also showed clear differences between the values of MPF and RMS amplitude for the signals collected at the start of each trial, depending on what abduction angle was being held, the gender of the subject and the arm from which the signals were obtained. It is important to stress that, given the way in which the beginning of the trials was structured, MPF and RMS amplitude from the first recorded signals actually reflect the muscles' response to the adoption of the experimental posture by the subject.

To assess the significance of the influence of the experimental factors, the 89 values of MPF and RMS amplitude at time zero were submitted to the statistical tests best suited to each case. Thus, two-sample t-tests were carried out on the values grouped according either to the arm from where the signals were obtained or to the gender of the subject, and the results are presented in tables 7.5 and 7.6, respectively. One-way analysis of variance with the abduction angle as test factor produced the results presented in table 7.7, which also includes the results of the Tukey test for multiple comparisons that was applied to probe into the differences between pairs of angles.

Table 7.5 Results from two-sample t-tests performed to assess the difference in the characteristics of the EMG signals collected at time = 0 from left and right arm.

	EMG Characteristic					
	Average Mean Power Frequency (Hz)			Average RMS amplitude (mV)		
	Muscle			Muscle		
Samples from arm	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
Right	67.1	77.8	69.9	607	610	382
Left	61.9	77.8	66,0	560	547	250
t value	4.3	0.08	3.38	1.43	1.43	7.38
p value	<0.001	N.S.	<0.001	N.S.	N.S.	<0.001

Table 7.5 shows that, with the exception of MPF from the medial deltoid where they were equal, the average starting value of both EMG features was larger for the muscles of the right arm than for those of the left arm. The differences were significant for MPF from the trapezius and the posterior deltoid muscles, but only the right posterior deltoid muscle had an average RMS amplitude at time zero larger than its counterpart in the left arm.

In relation with this last finding, it is pertinent to recall that earlier in the chapter (see section 7.3.3), the values of RMS amplitude were used to determine the average level of activation exhibited by the muscles throughout the trials. Now, a comparison of the information presented in tables 7.5 and 7.3 shows that the values of RMS amplitude for the trapezius and the posterior deltoid muscles of the right arm remained consistently larger than in the corresponding muscles of the left arm; in fact, the difference (right - left) increased sharply during the exertion and so it reached significance for the trapezius muscles ($z = 10.8$, $p < 0.001$) and grew even more substantial between the posterior deltoids ($z = 13.49$, $p < 0.001$). In contrast, the slight, non-significant difference between the medial deltoid muscles observed at the moment the subjects adopted the postures (holding time = 0) disappeared completely in the course of the trials ($z = -0.22$, $p > 0.05$). Also remarkable is the fact that for both muscles the values exhibited practically identical dispersion around the mean.

The gender of the subject also had an influence on the starting values of both EMG features, and it exhibited an interesting pattern, as shown in table 7.6: whilst the male subjects had higher values of MPF and RMS amplitude in both right and left trapezius muscles, the female subjects had higher values of both EMG features in the deltoid muscles of both arms. This difference between the genders was significant for the starting values of MPF from the medial and the posterior deltoid in the right arm, and from the trapezius and the medial deltoid in the left arm. The difference in RMS amplitude was significant only for the values from trapezius and posterior deltoid in the left arm. Thus, it appears that the influence of the subject's gender was stronger on the starting values of MPF than on those of RMS amplitude, with the values for the male subjects being larger than those of the females.

Table 7.6 Results from two-sample t-tests performed to assess the influence of the gender of the subject on the characteristics of the EMG signal collected at time = 0 throughout 89 trials of maximal holding of arm abduction.

	EMG Characteristic											
	Mean Power Frequency (Hz)						RMS Amplitude (mV)					
	Right shoulder			Left shoulder			Right shoulder			Left shoulder		
Subject's Gender	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
Female (45 trials)	65.5	79.6	72.4	60.2	81.6	67,0	591	632	402	466	587	274
Male (44 trials)	68.6	76.2	67.4	63.5	74.2	65.1	623	589	362	650	510	228
t value	-1.73	2.64	3.63	-2.13	5.95	1.07	-0.69	0.7	1.27	-4.35	1.24	2.82
p value	N.S.	<0.01	<0.001	<0.05	<0.001	N.S.	N.S.	N.S.	N.S.	<0.001	N.S.	<0.01

Table 7.7 Results from the analysis of variance performed to assess the influence of the abduction angle on the characteristics of the EMG signal collected at time = 0 throughout 89 trials of maximal holding of arm abduction.

	EMG Characteristic											
	Average Mean Power Frequency (Hz).						RMS Amplitude (mV) (Average over 89 trials)					
	Right shoulder			Left shoulder			Right shoulder			Left shoulder		
Angle	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
30°	63.4	75.5	66.9	57.8	74.5	62.5	541	402	345	482	390	224
60°	64.5	78.2	70,0	60.4	78.1	65.5	605	572	379	545	520	233
90°	73.7	79.7	72.8	67.6	80.7	70.1	676	859	422	654	732	295
F value	17.7	3.64	5.85	19.25	6.72	7.2	2.9 (n.s.)	31.28	1.99(n.s.)	4.91	12.92	8.62
Significant differences	90°>30° 90°>60°	90°>30°	90°>30°	90°>30° 90°>60°	90°>30°	90°>30°	None	60°>30° 90°>30° 90°>60°	None	90°>30°	90°>30°	90°>30° 90°>60°

7.4.3 Influence of the abduction angle on the EMG features at the start and throughout the holding

Table 7.7 shows that for all the six muscles the average value of MPF at the start of the trials was higher at the larger abduction angles. However, although all the F values were significant, the difference between the mean values of MPF was not significant for all pairs of angles. The average starting value of RMS amplitude also increased at the larger angles, but whilst the increase was not significant for either right trapezius or right posterior muscles, there was a significant difference for the right medial deltoid between every pair of angles.

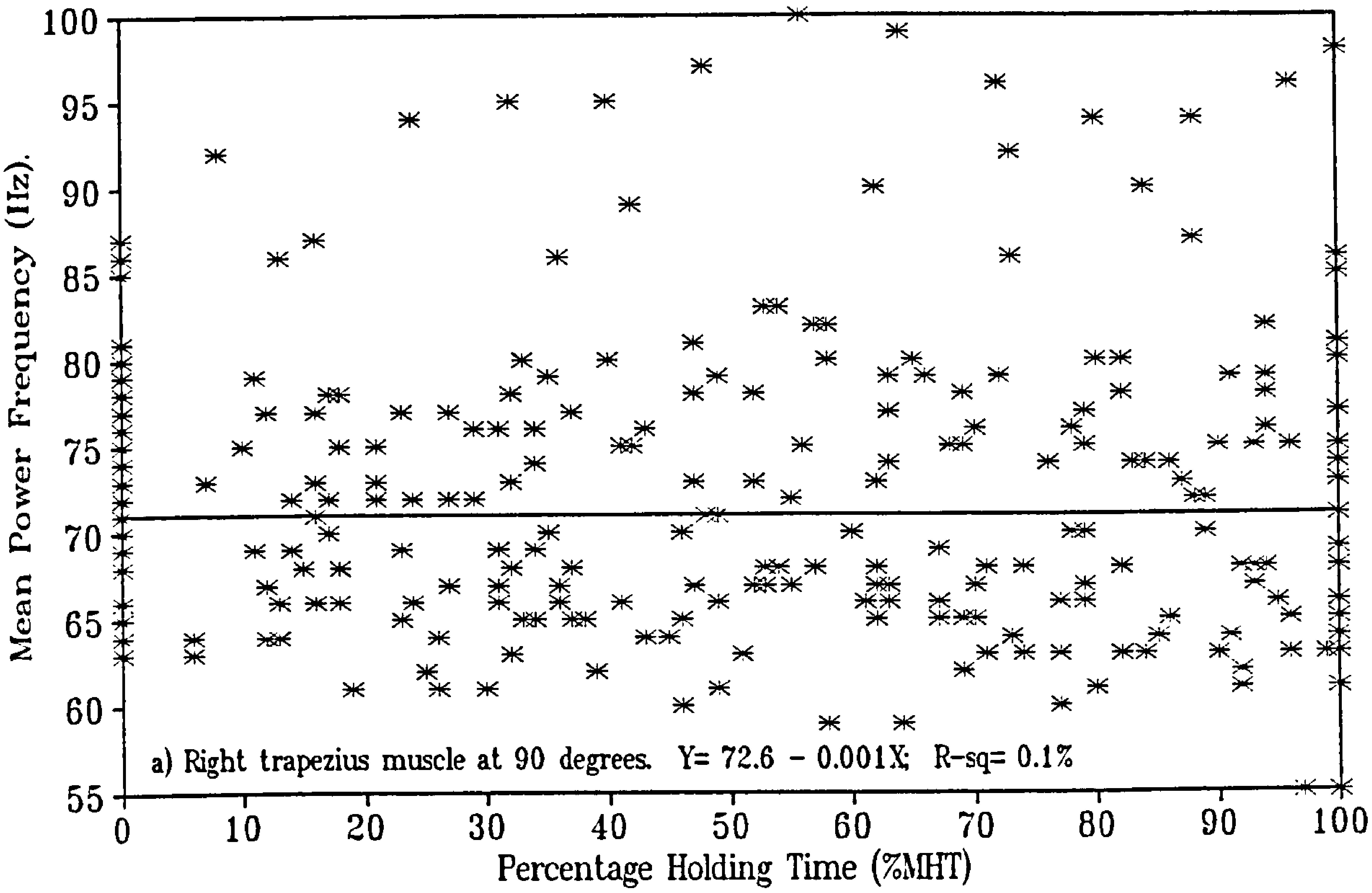
It has then been shown that the factor with the most consistent (and the strongest as well) influence on the values of MPF and RMS amplitude at the start of each of the 89 trials was the abduction angle. It might therefore be assumed that this factor would also affect considerably the features of the relationship between the values of the electromyographic parameters and the passage of the holding time. To assess the strength of this assumption, a regression model was fitted to the whole set of data calculated from the signals collected during the holding trials at each of the abduction angles. The results are presented in table 7.8.

Table 7.8 Parameters of the regression line fitted to the values of MPF and RMS amplitude over percentage holding time. The regression models were calculated for the sets of EMG signals collected during the 89 trials to measure MHT. (29 trials at 30°, 30 trials at 60°, 30 trials at 90°). The slope values marked with * were non significant.

	EMG Response											
	Mean Power Frequency (Hz)						RMS Amplitude (mV)					
Angle	Right shoulder			Left shoulder			Right shoulder			Left shoulder		
30°	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
Constant	63.8	70.3	62.3	55.6	69.3	59.1	507	355	299	420	316	196
Slope	-0.017*	-0.029	-0.036	0.012*	-0.016*	-0.024	2.44	1.93	0.89	3.45	1.55	0.58
R^2	0.7%	3.3%	2.2%	0.3%	0.6%	1.1%	12.1%	24.6%	3.2%	18.7%	20.5%	9.2%
60°	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
Constant	65.7	71.1	64.5	60.4	72.1	60.6	555	493	313	549	407	207
Slope	0.051	-0.05	-0.054	0.024*	-0.036	-0.039	1.97	1.94	1.35	4.39	1.73	1.26
R^2	2.2%	6.8%	5.8%	1.1%	3.0%	2.6%	9.3%	14.3%	9.9%	25.0%	10.2%	16.2%
90°	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
Constant	72.6	74.4	69.5	66.5	76.4	65.9	649	787	391	656	645	273
Slope	-0.001*	-0.181	-0.097	-0.017*	-0.152	-0.071	2.87	1.02*	2.49	4.53	1.85	2.36
R^2	0.1%	39.1%	17.7%	0.9%	30.9%	7.0%	9.1%	1.7%	16.1%	18.2%	4.7%	11.5%

The dominant feature of the information given in table 7.8 is the quite low values of R^2, a consequence of the dispersion of the data, which was more important for the values of MPF, as indicated by the number of non-significant slopes. Nonetheless, the values for MPF from both medial deltoids at 90° and for RMS amplitude in the same muscles, only this time at 30° were substantially higher than the rest. At 60°, only the value for RMS amplitude from the left trapezius was clearly above the rest. Thus, whilst these disparities in the value of R^2 show that the presence of a linear pattern of change in the EMG signals was more readily apparent in the medial deltoids, the fact that it was not found in the same EMG feature and, furthermore, it occurred at the two extremes of the abduction range investigated shows that the influence of this factor (the abduction angle) was not consistent. Figures 7.4 and 7.5 show the extreme examples of goodness of fit for both EMG parameters. The influence of the dispersion of the data on the goodness of fit of the regression models is also visible, particularly for those cases illustrated in figure 7.4.

On the other hand, the main aim behind the construction of table 7.8 was well justified by the information it contains, since it makes quite evident that the abduction angle had in fact an influence on the goodness of fit of the regression model. However, its effect on the six muscles studied did not exhibit a consistent pattern, and neither did it affect the two EMG features in a similar manner. This is best illustrated by looking at the value of R^2 for the regression models, particularly those fitted to the values calculated from the signals collected from both trapezius muscles.



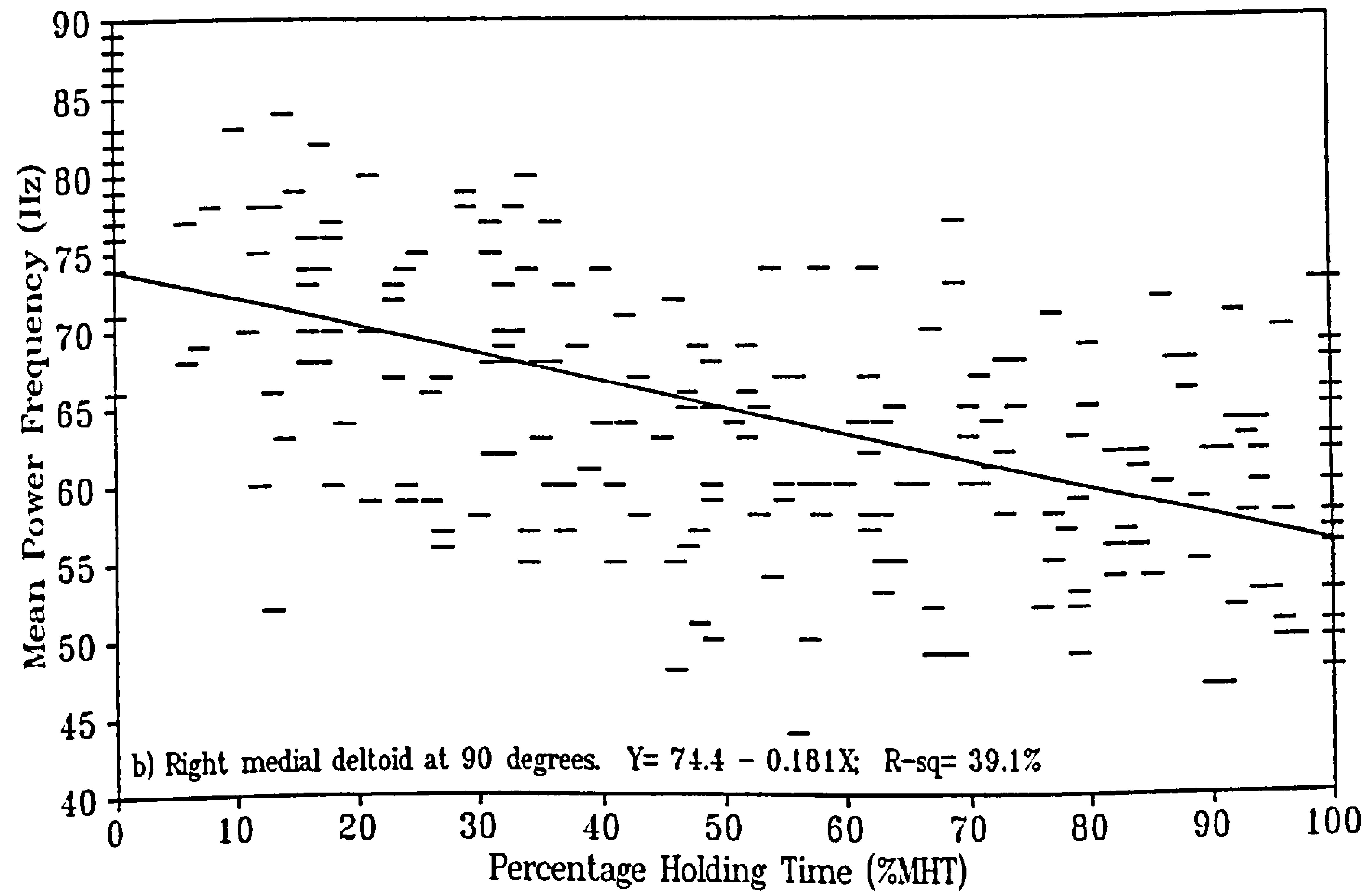
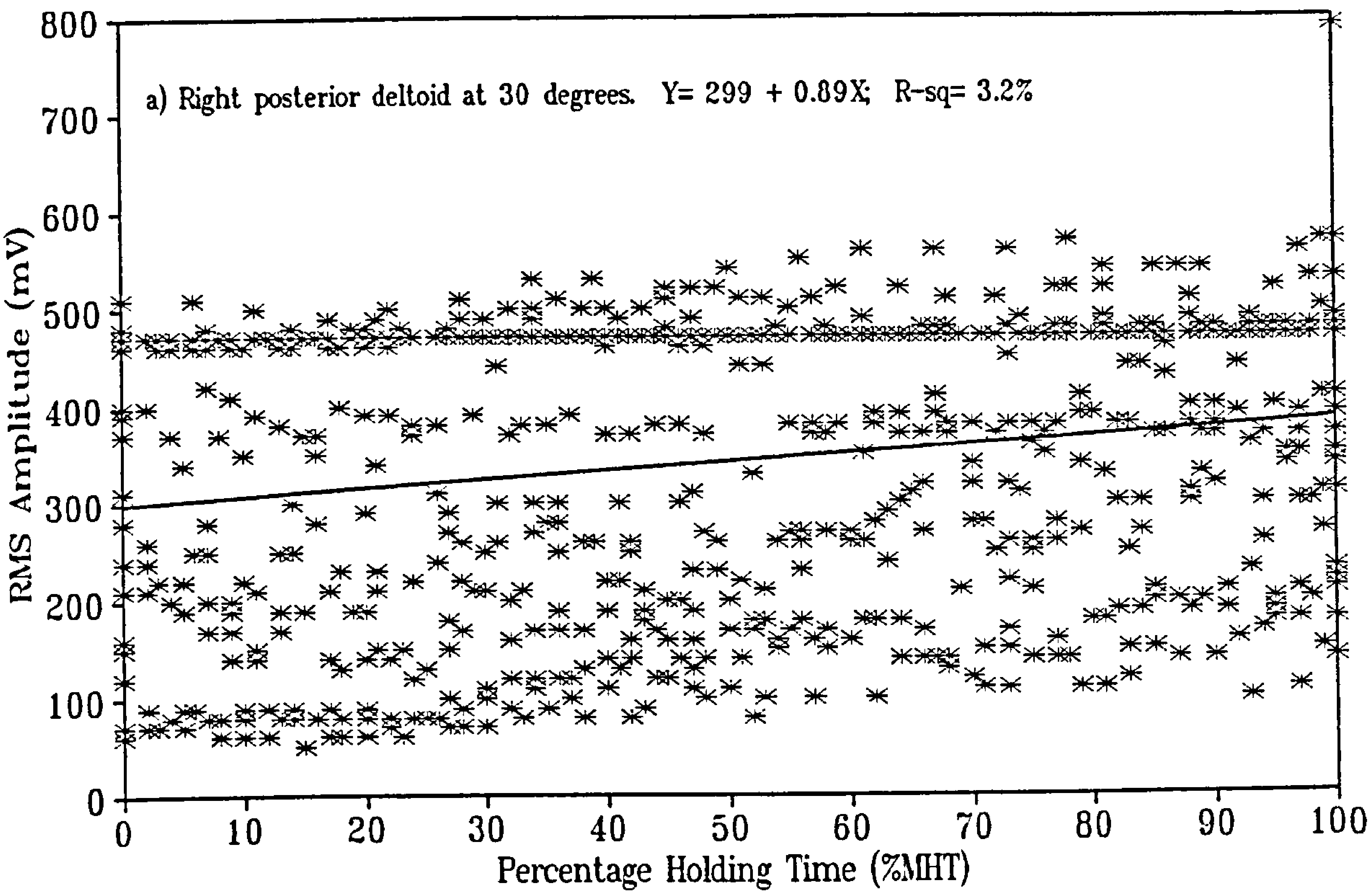


Figure 7.4 Regression line fitted to the values of MPF for the EMG signals collected during holdings to exhaustion of both arms abducted. a) Poorest fit; b) Best fit.



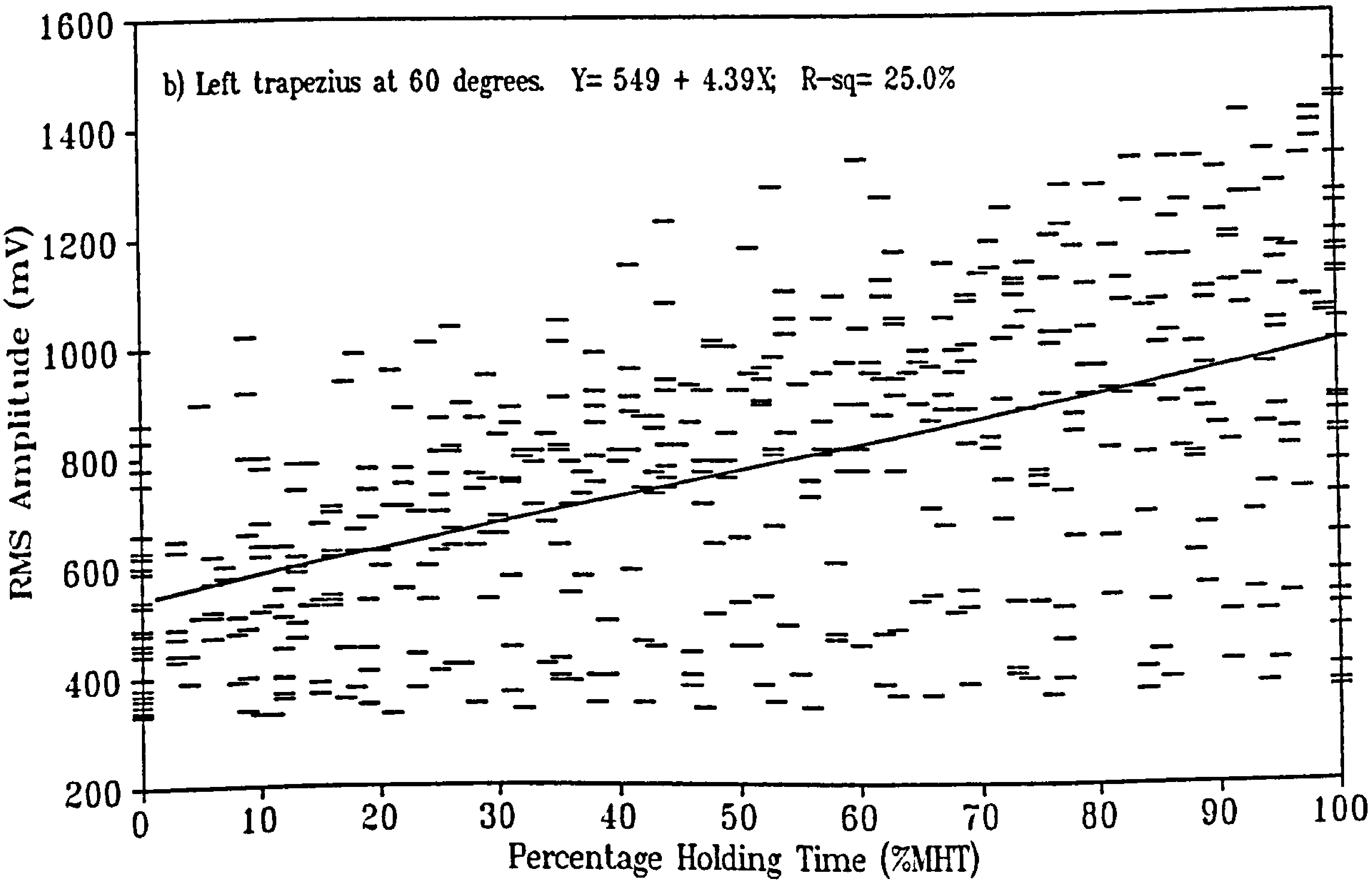


Figure 7.5 Regression line fitted to the values of RMS amplitude for the EMG signals collected durin holdings to exhaustion of both arms abducted. a) Poorest fit; b) Best fit.

However, the parameters of the models fitted to the values calculated for the EMG signals from the deltoid muscles (table 7.8) also had features that are worth a comment. Thus, whilst the coefficient of determination for the models using the values of MPF from both left and right medial deltoid showed a sharp improvement with the increase of the angle, those for RMS amplitude deteriorated, although the effect was less drastic. By contrast, the effects on the models for the posterior deltoid were more of a mixture, for in this case the increase of the angle meant an improved fit for the models using the MPF values of both arms and the RMS amplitude values of the right arm, but there was not a definite trend for the models fitted to the values of RMS amplitude from the left arm.

7.4.4 Extent of the changes in the EMG signal during the postural exertion

The changes in the characteristics of the EMG signal provoked by posture holding to exhaustion can be measured by calculating the differences in the values of MPF and RMS amplitude for every signal collected at the beginning an end of each trial. This difference was expressed as a percentage of the initial value, in order to express on a common basis the extent of the changes that occurred during the holding of the posture (all beginning at 0%), irrespective of the actual initial value of MPF and RMS amplitude which, as has just been shown in the last two sections, differed depending on the conditions of each trial, particularly the abduction angle. Appendix C contains the value of that difference for each of the six muscles studied, calculated for each of the 89 trials performed.

Table 7.9 presents the mean and standard deviation (given as mean ± s.d.) of the 89 differences for each muscle, calculated using both the actual values (where some changes were negative and some positive) and the absolute values of the differences.

Table 7.9 Mean and standard deviation of the percentage change in the EMG signal between the beginning and the end of the 89 trials of posture holding to exhaustion.

	Right Shoulder			Left Shoulder		
EMG feature	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
Actual MPF	0.2% ±10.93	-15.9% ±9.50	-12.1% ±9.44	-0.5% ±10.87	-13.5% ±11.28	-9.0% ±10.96
|MPF|	8.4% ±6.99	16.3% ±8.65	13.2% ±7.82	8.4% ±6.8	13.9% ±10.88	11.2% ±8.71
Actual RMS amplitude	38.7% ±48.94	24.1% ±42.62	55.0% ±90.66	67.7% ±62.17	37.0% ±58.63	50.4% ±81.99
|RMS amplitude|	38.9% ±48.73	31.6% ±37.35	55.3% ±90.5	68.0% ±61.77	43.7% ±53.75	52.5% ±80.85

The information contained in table 7.9 shows that for the six muscles studied, the average of the overall change in RMS amplitude was much larger than the corresponding average change in MPF, and that difference appears to have been more marked in the left arm. This table also shows that the presence of the reversed trend of change was the most noticeable in the values of MPF from both trapezius muscles. The effect of that trend was such that whilst the average of the actual values, which were affected by + or - sign depending on the nature of the change, was near zero (+0.22 % for the right trapezius, -0.55 for the left), the average of the absolute values was close to 8.5% in both arms. The change of RMS amplitude from the right medial deltoid was the only other variable evidently affected by the reversed trend; the presence of the reductions (the unexpected changes) meant a difference of 7.5% between the average of the actual values and the average of the absolute values.

7.4.5 The nature of the relationship between EMG changes and discomfort ratings

Up to this point, work has been completed on the first and second of the four subsidiary goals set for the collection of electromyographical data. The third of these goals will be tackled now, by linking the information obtained from the analysis so far performed on those data with the results of the study of the subjective perception of fatigue. This is done in order to establish if there is a significant relationship between these two indicators of fatigue and, were it so, its nature and strength.

To enable the search for the relationship between the change in the subjective perception of the fatigue (reflected in the increase of the discomfort ratings) and the changes in the features of the EMG signal, the difference between the values of MPF and RMS amplitude calculated from each of the signals collected during a trial and the corresponding value in the signal registered at time zero of that trial was converted into a percentage of this value. This procedure was applied in order to express those differences on a basis such that their extent was not affected by the differences between the actual values of MPF and RMS amplitude for the muscles, which existed both at the start of the trials and throughout them, and that have been considered already in sections 7.4.2 and 7.4.3. From now on, when reference is made to the percentage indices thus obtained, they will be called 'ΔMPF' and 'ΔRMS amplitude' when considered separately, 'ΔEMG' when referred to jointly.

It is convenient to recall that in all the 89 trials of maximum holding time performed, the subjective ratings of discomfort exhibited a very strong linear correlation with the holding time (see chapter 6). Thus, assessing whether those ratings were linearly related to the EMG changes was seen as an obvious approach. The assessment was performed by fitting a linear regression model to the data pairs constituted by the value of discomfort rating obtained at each sampling point and the corresponding ΔEMG index. The generic structure given to the regression models was $\Delta EMG = b0 \pm b1\ DR$, where DR stands for discomfort rating. The results of this procedure are presented in table 7.10.

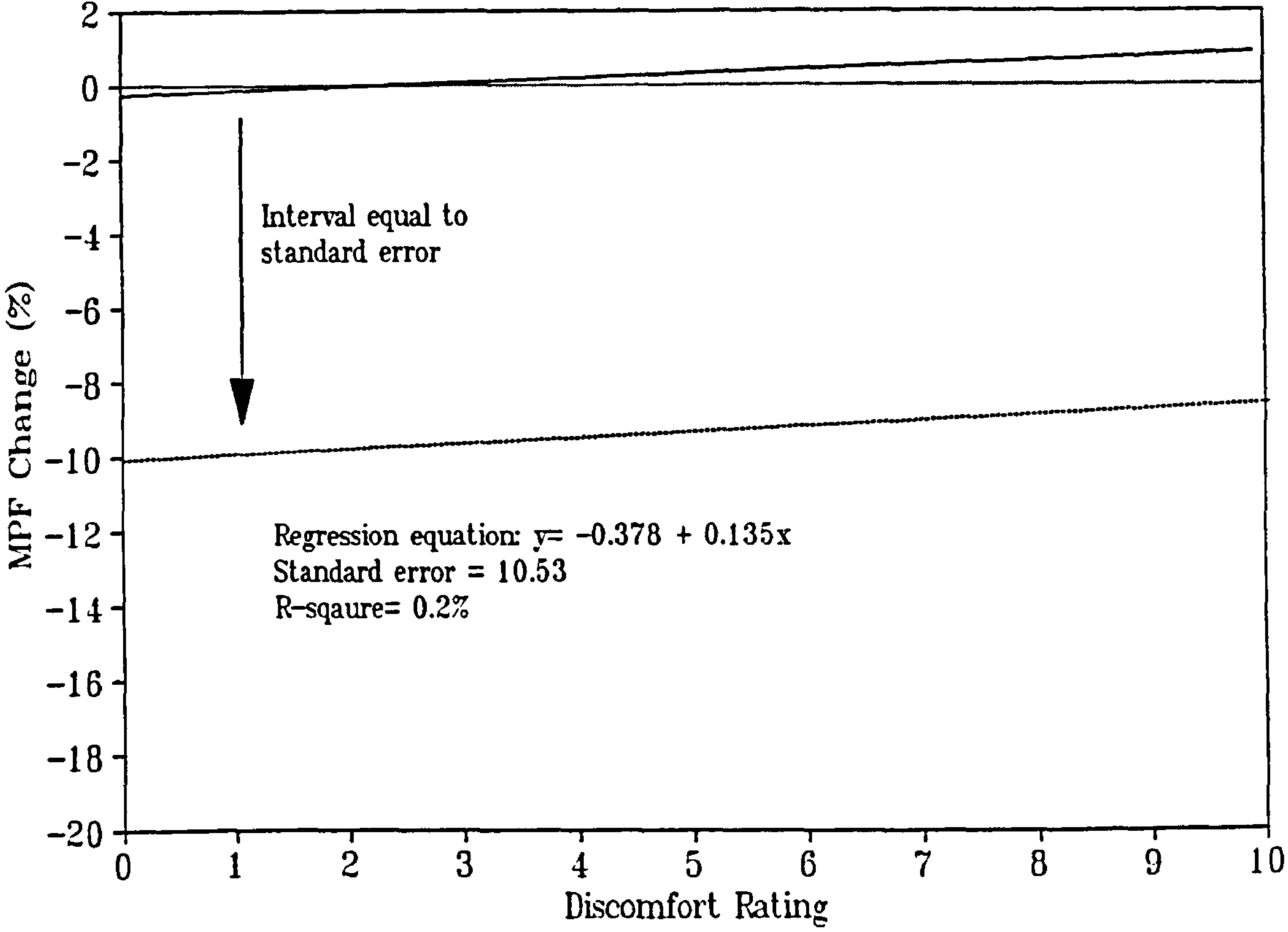
Table 7.10 Parameters of the regression model fitted to the data sets formed by a discomfort rating and a value of ΔEMG index. The regression models were calculated for the 1495 sets derived from the samples obtained during the 89 trials.

ΔEMG Index	Right Shoulder			Left Shoulder		
ΔMPF	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
Constant	- 0.378	- 6.75	- 5.64	- 2.47	- 6.26	- 5.32
Slope	0.14	- 0.726	- 0.601	0.19	- 0.531	- 0.54
R^2	0.2%	8.1%	4.8%	0.7%	4.5%	4.6%
F* value	2.95	131.29	73.73	9.96	69.61	70.45
p value	N.S.	<0.001	<0.001	<0.01	<0.001	<0.001
ΔRMS Amplitude	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
Constant	1.73	- 1.49	- 0.693	0.9	- 0.947	- 0.755
Slope	1.59	0.94	0.68	2.38	1.21	0.7
R^2	22.5%	15.4%	14.9%	33.7%	16.4%	13.0%
F* value	429.15	269.75	258.84	755.84	287.99	219.84
p value	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001

Were it interpreted only on the basis of the values of the F* statistic and the level of significance indicated by the p-value, the information presented in table 7.10 would lead to conclude that only the change in MPF from the right trapezius muscle failed to exhibit a significant linear relationship with the change in the discomfort ratings. However, when the values of R^2 were considered, they painted a very different picture altogether. Whilst it is true that they indicated the existence of a linear relationship between ΔRMS amplitude and discomfort ratings, whose strength may be described as moderate in the best case (that of the left trapezius, with $R^2 = 33.7\%$), they also made quite evident that for ΔMPF, despite some quite impressive values of F*, there is practically no linear relationship with the discomfort ratings for any of the six muscles. The sharpest contrast in the goodness of fit of the regression model was between ΔMPF for the right trapezius muscle ($R^2 = 0.2\%$) and ΔRMS amplitude for the left trapezius ($R^2 = 33.7\%$). Figure 7.6 illustrates such contrast by depicting the regression line defined by each of these models.

Thus, the conclusion that may be drawn from the results reported here is that, despite the remarkably strong linear relationship between the discomfort ratings and the holding time, this did not translate into a relationship of a similar kind with the indices of electromyographic change. It may be hypothesised at this stage that this was due to the erratic behaviour of the EMG signals which has been extensively reviewed already. Regrettably, time pressures prevented the possibility of exploring in more detail this issue, but this is without any doubt a very relevant issue that deserves more investigation, since it is the key to establishing meaningful links between the physiological phenomena and the responses they elicit from the psyche.

a) MPF Change in Right Trapezius
Regression Against Discomfort Rating



b) RMS Change in Left Trapezius
Regression Against Discomfort Rating

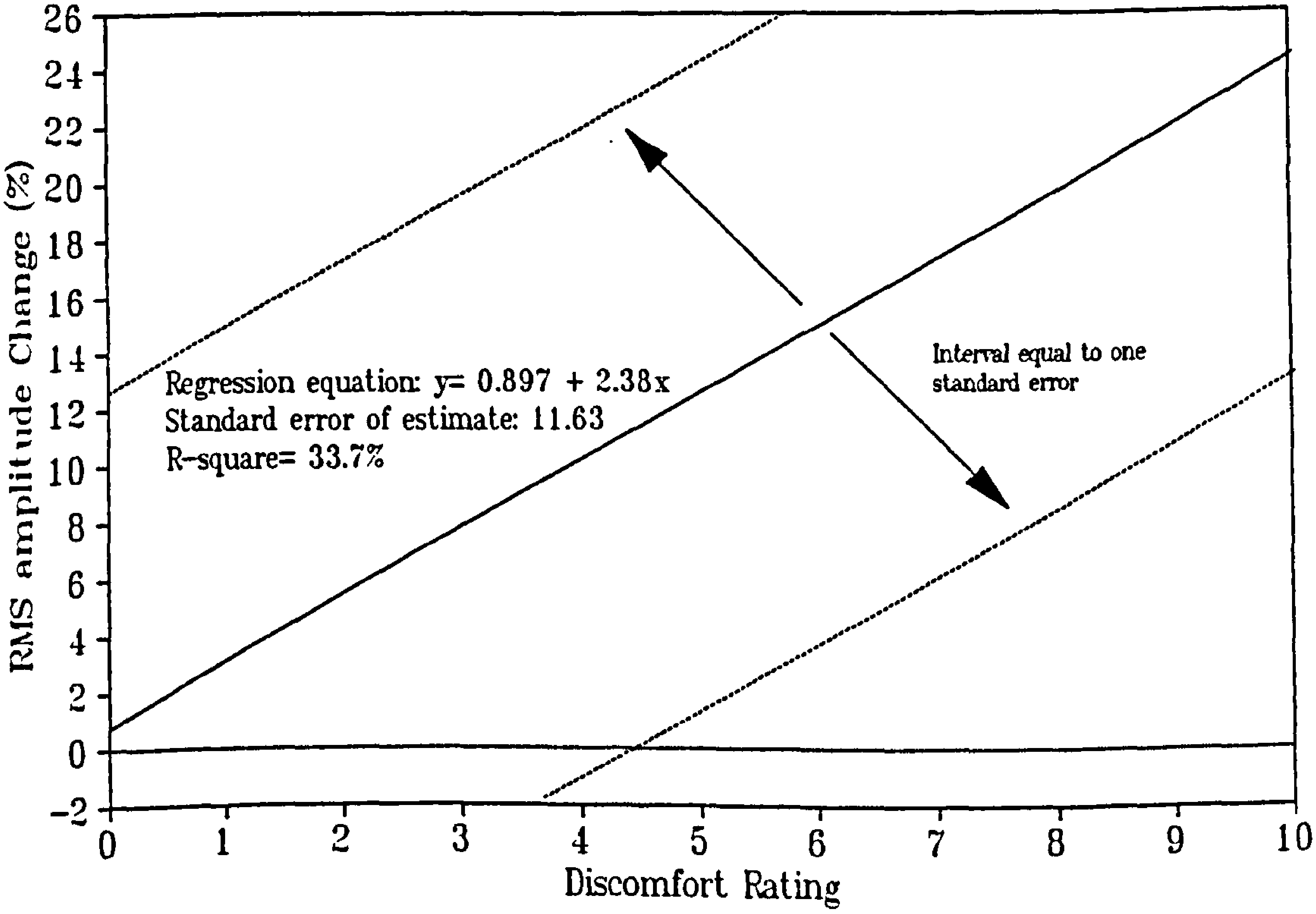


Figure 7.6 Extreme examples of the strength of the relationship between EMG changes and discomfort ratings.

7.5 Discussion

7.5.1 Consistency of the experimental conditions

The initial inspection of the results revealed two rather unexpected features in the change of the EMG signals that occurred during the holding to the endurance limit of postural exertion:

a) in many cases it did not follow a smooth, well defined pattern, and this appeared to be related to the length of the trial, being less well-defined in the less stressful postures which could be held for a long time;

b) especially during the trials at 60°, the changes in MPF for the trapezius muscles and the changes in RMS amplitude for the medial deltoid muscles often went in a direction opposite to that expected.

However, before discussing the relevance of the changes in the EMG signals as indicators of muscular fatigue, it is convenient to consider the possibility of those two occurrences being due to unintended variations in the experimental procedure, particularly during the collection of the EMG signals, since it has been demonstrated that the characteristics of the signal collected through surface electrodes may be greatly affected by even small variations in a number of factors, such as the positioning of the electrodes, the distance between them, and the resistance opposed by the skin and subjacent tissues (Basmajian and De Luca, 1985, De Luca, 1985, Veiersted, 1991).

To test the consistency of the set-up for the collection of the EMG signals throughout the 89 trials, the values of MPF and RMS amplitude of the signals collected during the reference muscular contraction performed before each trial were submitted to analysis of variance, with the repetition at each abduction angle as the test factor. The results showed no significant difference between the values of MPF (F values ranged between 0.06, $p>0.9$ and 0.78, $p>0.5$) or between those of RMS amplitude (F ranged from 0.01, $p>0.99$ to 0.93, $p>0.4$). The signals collected at the beginning of each trial (at time = 0) were analysed in the same way, and again there were no significant differences: when testing on MPF the value of F ranged from 0.01 ($p>0.99$) to 1.43 ($p>0.2$) and the tests on RMS amplitude yielded F values between 0.14 ($p>0.8$) and 0.88 ($p>0.4$). These results confirm that the unusual features observed in the change of the EMG signal were not attributable to variations in the experimental set-up, but are a true reflection of the way the muscles behaved during the experiments.

7.5.2 EMG changes as indicators of muscular fatigue

The presence of a change in the characteristics of the EMG signal, between the start and the end of a muscular effort, has often been used as the evidence of the existence of fatigue. However, the application of such criteria to the changes that occurred during the holding of the postures could possibly yield misleading results, for it would concentrate only on two values at start and end of the effort, without taking into consideration the large variations exhibited by the features of the EMG signal in the course of many of the trials.

The plots of the change in the EMG signal over the holding time (included in appendix D) show that in many trials, although there were considerable changes between consecutive sampling points, these frequently went in opposing directions and tended to cancel each other, resulting in an end value of MPF or RMS amplitude which was not very different from the initial one, making the difference between beginning and end of the trial appear almost negligible. Further evidence of such cancelling effect comes from the values of the slope coefficient for the linear regression models presented in table 7.8: the majority of them were very close to zero, showing that the values of the corresponding variable tended to spread rather evenly around a central value that was close to the initial one.

To account for the variations in the EMG features between consecutive sampling points, the percentage change at each point with respect to the initial value was calculated, and these data were then submitted to a t-test to evaluate whether the change throughout the trial was significantly different from zero. If that were the case, it would be the evidence of the presence of muscular fatigue. Table 7.11 presents the number of trials for which the t-test showed that the average percentage change in MPF and RMS amplitude was significantly different from zero.

Table 7.11 Number, out of 89 trials of posture holding to exhaustion, in which the average change in the characteristics of the EMG signal throughout the trial was significantly different from zero.

	Right Shoulder			Left Shoulder		
EMG feature	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
MPF	66	87	77	62	80	74
RMS amplitude	73	64	56	79	65	65

The information contained in table 7.11 shows that, for the six muscles studied, the great majority of the 89 trials resulted in changes to the characteristics of the EMG signal whose average was significantly different from zero. Therefore, muscular fatigue did occur as a result of pure postural exertion, and it was evidenced by changes in the electromyographic activity. Indeed, as shown in table 7.9, in the six muscles studied, the average absolute value of change in MPF between the start and the end of the experiments was larger than 8%, a size of change that Öberg et al (1990) considered to be an unequivocal indicator of the presence of localised muscular fatigue. Apparently no similar threshold has been proposed for the change in RMS amplitude, but the size of the change found in the present study ranged between 31% (in the right medial deltoid) and 68% (left trapezius).

Another important fact borne out by the information included in table 7.11 was that, apparently, the two measures derived from the EMG signal were not equally well suited to indicate, in each of the six muscles, the presence of fatigue through an average change significantly different from zero. Thus, for both the medial and the posterior deltoid muscles, the mean change of MPF was significantly different from zero in more cases than it was for RMS amplitude, but for the trapezius muscles the situation was reversed. This finding could have important implications, since it would mean that different muscles would require different modes of evaluation to determine whether or not fatigue is occurring. This issue will be treated at length later, in the Discussion chapter.

7.5.3 Direction of the changes in the EMG parameters

Among the 89 trials of posture holding to exhaustion, there was a considerable number of cases where the change in the EMG signal went in direction opposite to that expected, since MPF increased instead of decreasing and RMS amplitude decreased instead of increasing.

Information regarding the number of trials in which either of those reversed changes occurred was given in table 7.2. Those numbers were analysed to search whether any of the experimental factors was linked to this phenomenon, but apart from the fact that the reversed change of MPF was found more frequently in the trapezius muscles, and the preferred site for the unexpected increase of RMS amplitude was the medial deltoid, no other evident link could be established. The probable causes for the reversed EMG changes, and their effect on the evaluation of muscular fatigue will be reviewed in the Discussion chapter.

7.5.4 Influence of the main experimental factors on the electromyographic changes

Having established that the postural exertion does in fact provoke muscular fatigue, and does produce changes in the EMG signals, attention can be turned to the possible influence that the experimental conditions, namely gender of the subject and abduction angle, may have exerted on the nature and extent of the EMG responses.

7.5.4.1 Influence of the gender of the subject

For a start, there are two aspects of the EMG changes in which it is worth investigating the influence of the gender of the subject: 1) the number of trials executed by the subjects of each gender in which a significant change occurred and 2) the extent of those changes.

To look into the first of those aspects, the number of significant EMG changes that occurred in each muscle was classified according to the gender of the subject who presented it, these subtotals were then converted into a proportion of the total number of trials performed by each group and tested for significant differences. Thus, except for MPF in the right trapezius, the proportion of significant changes in MPF and RMS amplitude was larger for the male subjects than it was for the females, only slightly larger for MPF, but much larger for RMS amplitude. When those differences were tested for significance, it emerged that in fact more significant changes of MPF in the left trapezius and of RMS amplitude in the three muscles of the right arm plus left trapezius occurred among the male subjects. Thus, in all appearance it was true that more significant changes of RMS amplitude did occur among the males subjects, although it is not easily apparent why it should happen more often in the muscles of the left arm.

To test whether the extent of the changes in the EMG signal was influenced by the gender of the subject, the percentage changes observed throughout the 89 trials were submitted to one-way analysis of variance, with the gender as test factor. The results of this procedure are presented in table 7.12. It shows that the gender of the subject had a significant influence on the extent of the average percentage change of RMS amplitude in the six muscles, and in all the cases the change was larger among the male subjects. Regarding the average percentage change of MPF, the gender of the subject appeared to affect it significantly only on three muscles, and again the trend was towards larger changes for the male subjects.

Table 7.12 Results from the analysis of variance carried out to assess the influence of the gender of the subject on the extent of the EMG response to arm abduction.

	Average percentage Change in EMG Response											
	Mean Power Frequency						RMS Amplitude					
	Right shoulder			Left shoulder			Right shoulder			Left shoulder		
Subject's Gender	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
Female	0.59%	-10.65%	-7.59%	0.30%	-11.36%	-8.13%	12.19%	11.47%	5.04%	31.64%	1.05%	7.75%
Male	0.10%	-10.11%	-9.15%	-2.60%	-7.43%	-7.86%	32.28%	14.09%	42.33%	43.23%	36.74%	24.81%
F value	0.75	1.29	9.08	45,00	73.15	0.31	118.37	23.49	147.39	19.68	276.06	48.69
p value	N.S.	N.S.	<0.001	<0.01	<0.001	N.S.	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001

However, in interpreting the information provided by table 7.12 it needs to be considered that the gender of the subject may not actually be the underlying cause for some of the differences demonstrated by the statistical analysis, since the data were also affected by the presence of the trials with reversed changes of the EMG features. This appears to be particularly the case with ΔMPF for both genders in the left trapezius and with ΔRMS amplitude in the left medial deltoid for the female subjects. If these three cases were disregarded, then the conclusion would be that the gender of the subject was in fact linked with significant differences in the extent of the EMG changes, and its association was particularly strong with RMS amplitude.

7.5.4.2 Influence of the abduction angle

Breaking down the number of trials with significant changes, in function of the abduction angle, was again the first step. The result is presented in table 7.13. It shows that the abduction angle did not appear to have any systematic influence on the number of trials with a significant change in either of the EMG features. This was reflected by the outcome of the statistical test applied (χ^2 test): for the changes in MPF the result was $\chi^2=1.964$, $p>0.25$, and for RMS amplitude it was $\chi^2=6.450$, $p>0.05$.

Table 7.13 Breakdown, by abduction angle, of the number of trials of posture holding to exhaustion with average change in the characteristics of the EMG signal significantly different from zero

	Trials with significant change in EMG response											
	Mean Power Frequency						RMS Amplitude					
	Right shoulder			Left shoulder			Right shoulder			Left shoulder		
Angle	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
30°	22	29	27	22	25	27	26	26	16	27	23	19
60°	24	30	28	18	27	24	25	25	19	27	26	25
90°	20	28	22	22	28	23	22	13	21	25	16	21
Total	66	87	77	62	80	74	73	64	56	79	65	65

The outcome of the one-way analysis of variance on the average percentage change throughout the total number of trials carried out at each of the three abduction angles is presented in table 7.14; the difference between pairs of angles was evaluated by means of Tukey tests for multiple comparisons. This information suggests that the abduction angle appeared to affect the change of MPF in each of the six muscles studied in rather mixed ways. Only the two medial deltoid muscles exhibited what might be seen as the expected effect: the greater the abduction, the larger the average change, with the increase from one angle to the next being statistically significant, hence the high level of significance reached by the corresponding ANOVA tests. For the right posterior deltoid muscle, the mean change of MPF at 30° was significantly smaller than at 60° and 90°, but there was no difference between these two. For the left posterior deltoid, the difference was significant only between 30° and 90°.

Table 7.14 Results from the analysis of variance carried out to assess the influence of the abduction angle on the extent of the EMG response to arm abduction.

	Change in EMG Response											
	Mean Power Frequency						RMS Amplitude					
	Right shoulder			Left shoulder			Right shoulder			Left shoulder		
Angle	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
30°	-2.9%	-7.4%	-7.8%	-3.3%	-6.9%	-7.5%	28.5%	24.3%	20.9%	35.6%	25.3%	5.3%
60°	5.6%	-10.2%	-9.1%	1.4%	-8.4%	-7.8%	18.3%	14.2%	36.2%	43.0%	24.2%	28.0%
90°	-1.3%	-18.5%	-9.8%	-2.4%	-15.1%	-9.8%	27.5%	3.4%	35.2%	40.4%	18.7%	37.3%
F value	115.16	187.21	5.64	54.52	97.17	6.81	13.2	34.86	11.8	3.59	2.32	67.66
p value	<0.001	<0.001	<0.01	<0.001	<0.001	<0.01	<0.001	<0.001	<0.001	<0.05	N.S.	<0.001
Significant differences	60°>30° 60°>90°	60°>30° 90°>30° 90°>60°	60°>30° 90°>30°	60°>30° 60°>90°	60°>30° 90°>30° 90°>60°	90°>30°	30°>60° 90°>60°	30°>60° 30°>90° 60°>90°	60°>30° 90°>30°	60°>30° 90°>30°	None	60°>30° 90°>30°

The change in MPF from both trapezius muscles, however, deserves further comment. It has been mentioned already that the trend for MPF to increase instead of to decrease appeared more often in both trapezius muscles than in the other four muscles. Table 7.14 shows that this fact was reflected by the value of the average percentage change at the three abduction angles. Clearly, the strongest effect of the reversed trend of change was on the trials at 60°, whose average change was positive instead of negative, and this no doubt contributed in good measure to the high significance of the ANOVA test, particularly for the right arm. The results from the trials at the abduction angles of 30° and 90° also showed the effects of the reversed trend, in that although the average percentage change was negative, it was noticeably of a lesser magnitude.

The average percentage change of RMS amplitude exhibited even more varied responses to the increase of the abduction angle than did MPF. Thus, only the left posterior deltoid presented the expected changes, with the average value increasing significantly in response to each increase of the angle. In the right posterior deltoid, the value increased significantly with the passage from 30° to 60°, but the increase to 90° actually resulted in a slight decrease (1%) of the average change.

The ANOVA test (table 7.14) was significant for both trapezius muscles, at a higher level for the right arm. However, the changes did not follow the expected pattern of increase with the larger abduction angles, nor were all the differences between pairs of angles significant. In the right trapezius, the change at 60° was significantly smaller than at 30° and 90°, but the difference between these two was only 1%. In the left trapezius, by contrast, the average change increased significantly with the passage from 30° to 60°, and then dropped slightly when the angle increased to 90°, but there was no significant difference with the other two angles.

In both right and left medial deltoids, the average change of RMS amplitude went in direction opposite to what was expected, that is, it actually decreased at the larger abduction angles. However, whilst in the right arm the difference between every pair of angles was significant, resulting in a highly significant ANOVA test, in the left arm the differences were small (less than 1% between 30° and 60°), rendering the ANOVA test non-significant.

The downwards shift of RMS amplitude in the medial deltoids was not unexpected, since these were precisely the muscles where the reversed trend of change was the most frequent. However, the results from the ANOVA tests suggest that the effect was more evident in the EMG signals collected from the right arm during the trials at 60° and 90°, whilst it appeared to affect evenly the signals collected from the left arm at the three abduction angles.

The likely causes for the presence of the reversed changes in the EMG features, and its implications for the interpretation of the fatigue process in the muscles in which they appeared will be treated at length in the Discussion chapter.

7.5.5 Persistence of EMG signs of muscular fatigue after posture holding

As already mentioned in section 7.2.4, the persistence of the signs of muscular fatigue after a rest period was assessed by comparing the values of MPF and RMS amplitude calculated from the EMG signals collected during two reference muscular contractions (RMC). The first of these (RMC_1) was performed before the start of each trial, the second one (RMC_2) after the subject had rested for five minutes following the end of that trial. For each muscle, the value of MPF and RMS amplitude in the signal from RMC_2 was subtracted from the corresponding value in the signal from RMC_1, and the difference converted into a percentage of the latter. To test whether the five minutes' rest was enough to restore the EMG signal to the state it was in before the trial, the 89 differences $RMC_2 - RMC_1$ (one per trial) were put together and submitted to a t-test, probing for a significant departure from zero. Table 7.15 shows the results of these t-tests, together with the average value of the 89 differences on which they were performed.

Table 7.15 Results of the t-test on the average value of the difference between the EMG signals collected in the first (RMC$_1$) and second (RMC$_2$) reference muscular contractions during each trial. The difference is given as % of RMC$_1$

	Right Shoulder			Left Shoulder		
MPF	Trapezius	Medial Deltoid	Posterior Deltoid	Trapezius	Medial Deltoid	Posterior Deltoid
Mean ± s.d. of difference	-2.5% ±8.74	- 4.8 % ±8.77	-3.6 % ±9.07	-1.8 % ±7.45	-2.1 % ±10.59	-3.5 % ±8.14
t value	-2.62	-4.93	-3.38	-2.03	-0.91	-3.65
p value	<0.01	<0.001	<0.01	<0.05	N.S.	<0.001
RMS amplitude	Trapezius	Medial deltoid	Posterior deltoid	Trapezius	Medial deltoid	Posterior deltoid
Mean ± s.d. of difference	-3.4 % ±20.77	-1.0 % ±8.33	-7.4 % ±34.49	-1.8 % ±14.55	-3.5 % ±12.93	-1.8 % ±14.00
t value	-0.55	-1.36	-3.17	-2.72	-2.48	-1.1
p value	N.S.	N.S.	<0.01	<0.01	<0.05	N.S.

The fact that the average difference was negative in all the cases indicates that, following the rest period allowed, the EMG parameters had not only returned to the value they exhibited prior to the exertion, but were larger even.

These findings agree with those of Petrofsky and Lind (1980), Mills (1982), Merletti et al (1983) and Kuorinka (1988), all of whom have reported that, following the termination of a sustained isometric contraction in which the frequency parameter of the EMG signals (either MPF or median frequency) showed a significant downward shift, it returned to well within its initial value in a time span of between 3 and 5 minutes. These authors also observed that the return of frequency parameter to its initial value tended to follow an exponential pattern, with a short phase (1 minute according to Petrofsky and Lind, 1980; 1 to 3 minutes in Kuorinka, 1988) that exhibited a large return rate occurring immediately after the cessation of the exertion, and a second, slower phase, taking up to the fifth minute.

The phenomenon in which the value of the frequency parameter (median frequency in this case) in the measurements post-exertion went beyond its pre-exertion value was reported only by Merletti et al (1983), who referred to it as 'overshooting'. They interpreted its presence in the context of their experiments, which involved EMG recording from the first dorsal interosseus muscle during trials in which pairs of constant force abduction of the index finger were performed, first at 20% MVC and then at 80% MVC. The trials were conducted separately in conditions of ischaemia and of intramuscular cooling. The overshooting of the median frequency appeared only in connection with the conditions of restricted blood flow, and was attributed by Merletti et al (1983) to the increase of the intramuscular temperature and of metabolites clearance rate that occurred following the release of the blood blockage.

Considering the obvious differences between the experimental conditions applied in the present investigation (holding of a posture which involved the activation of large muscle groups to a low level of strength) and those in the experiment of Merletti et al (1983) described above, it is difficult to visualise how the explanation they offered for the presence of overshooting could apply to the results obtained during the main experiment. Therefore, the presence of overshooting not only in the values of MPF but also in those of RMS amplitude (which in one particular case reached almost 100%) will have to remain unexplained for the time being, and this no doubt opens a wide invitation for further research into the issue.

7.6 Conclusions

This chapter opened with the advancement of four objectives. Therefore, such is also the number of main conclusions that have been reached after reviewing the results of the study of the EMG information collected during the main experiment. These are:

i) The holding to exhaustion of the postural loads created by the abduction by men and women of both arms at 30°, 60° and 90° provoked changes in the myoelectrical activity which clearly indicate the presence of fatigue.

ii) Even though the extent of those changes was affected both by the gender of the subject and by the abduction angle, those effects did not follow a well-defined pattern.

iii) Evaluated on the overall data pool built during 89 trials, the relationship between the EMG changes and the subjective perception of discomfort was found not to be linear in nature.

iv) The signs of fatigue in the EMG signals disappeared following rest for five minutes.

CHAPTER 8

GENERAL DISCUSSION

8.1 Introduction

Each of chapters 4 to 7, which presented the experimental work carried out in this study, included a discussion section where the corresponding results were briefly reviewed and their relevance to the specific topic treated in that chapter was assessed. In this chapter, the whole of the results of the experimental work are reviewed together, considering the way they relate to the basic phenomena associated with the development of muscular fatigue, showing how they relate to the findings from other studies, evaluating their significance for the fulfilment of the objectives of this investigation and, as a corollary to this review, indicating those areas where the findings of this investigation showed the need (or convenience) of having more work done.

The discussion is divided into four main sections. The first reviews the non-applicability of Milner's model to the development of fatigue caused by a shoulder-loading posture, and considers the most likely reasons. Since Milner's model relies entirely on the maximum holding time (MHT) as the measure of endurance to postural loading, the consistency of such measure was seen as one of the most influential of those factors. Therefore, experimental work was undertaken to probe into this matter. The results from such experiment are considered in the second section.

All the experimental work of this investigation hinged around the maximum time that a posture may be held. The end point of the holding trials was marked by the subject, by reaching what they considered to be the limit to their capacity to endure discomfort. Therefore, another variable also considered in depth was the subjective perception of the growth of discomfort, and its relationship with the holding time. Such will be the matter of the third main section of this general discussion.

It was expected that the sustained holding of the posture would provoke changes in the muscular function which are widely taken to indicate the existence of fatigue. The collection and analysis of myoelectrical activity was the method used to measure the extent of those changes. The fourth section of the discussion will deal with the main issues related to such measurements and their usefulness as indicators of fatigue in purely postural exertion.

8.2 THE NON-APPLICABILITY OF MILNER'S MODEL FOR THE PREDICTION OF RECOVERY IN POSTURE HOLDING

The results of the trials with a sequence of two holdings of arms abduction at 60° were described in chapter 4. They showed that the model proposed by Milner (1985), with a view to predicting the extent of recovery to be expected in case of purely postural exertion, is not applicable to a posture different from the stoop from which it was developed. For the posture used in the present study, in upright position with abduction of both arms, the model tends to significantly overestimate the degree of recovery, defined as the length of the second holding until the subject reached the limit of endurance.

8.2.1 Possible reasons for the non-applicability of Milner's model

The discussion advanced in chapter 4 concentrated mainly on the possibility of improving the viability of the model by modifying its mathematical makeup. Now, consideration will be given to a number of issues related with the foundations of Milner's approach to the development of the model, assessing how realistic was his assertion about the transportability of the model to practically any posture.

8.2.1.1 Differences between the postures

It is quite obvious that the posture studied by Milner and the one studied in the present investigation differed substantially in regards of the body sites on which they place the main stress. Indeed, such difference was the main reason for wanting to test Milner's model on an upright posture with arms abducted. However, the different location of the most stressed sites may also mean a substantial difference in the anatomical structures most heavily involved in the bearing of the loads.

The stoop investigated by Milner was fairly close to the posture described in Rohmert et al (1986) as 'hanging by the ligaments', which involves a deep bend forwards with the arms hanging close to the floor. This posture was taken up by those authors in a study of the physiological and psychological effects of holding postures for which the discomfort and eventual fatigue would derive from a variety of anatomical structures. In the posture in question, those structures were assumed to be the tendons and ligaments of the lower back and the upper legs. On the other hand (almost literally), in arm abduction the discomfort is more likely to stem from the effort placed on the muscles of the shoulder and neck (Bjelle et al, 1979, 1981; Christensen, 1986; Jørgensen et al, 1989; Jensen et al, 1993).

Accordingly, those differences were the first factor that needed to be considered when looking for reasons to explain the failure of Milner's model to account for the results obtained in the present study. The relevance of this consideration was made clearly evident by the comments received during the presentation and discussion of the results from the early part of the present investigation (Serratos-Pérez and Haslegrave, 1992), when it was suggested that the disparity of the load-bearing structures was so crucial as to render any comparison between the two postures meaningless.

However, even though Milner defined the posture he tested very clearly, the issue could not be settled by resorting to evidence provided by his study, because despite having identified which were the body parts most affected by the discomfort created by the posture, Milner had no means of measuring the effect directly on the structures most likely to be the precise site where fatigue would develop. The evidence had to be gathered by this researcher, by carrying out a very short study to measure the muscle activity elicited by Milner's posture.

In such study, three male subjects who were not involved in the abduction trials were asked to adopt the posture tested by Milner and hold it to their limit of endurance. EMG signals were collected from right and left paravertebral muscles (at L_3 level), which have been shown to be heavily activated during forwards bending (Kippers and Parker, 1984; van Dieën et al, 1993). Besides, the lower back was one of the sites where Milner's subjects reported to have experienced heavy discomfort. EMG signals were also collected from right and left biceps femoris and internal gastrocnemius, since the thighs and the calves were also reported to grow very uncomfortable whilst holding Milner's posture. The presence of muscular fatigue was assessed by looking for significant changes ($p<0.05$) in the mean power frequency and RMS amplitude of the EMG signals.

Fatigue was clearly evident in the changes of MPF: the three subjects exhibited highly significant change ($p<0.001$) in both biceps femoris muscles, and significant change ($p<0.05$) in the right paravertebral and gastrocnemius muscles only. The changes in RMS amplitude were less homogeneous: one of the subjects showed a significant change in the left biceps femoris only, the second one showed it in the right biceps femoris, and the third subject had significant changes in the right gastrocnemius and the left paravertebral and biceps femoris. These results proved that Milner's posture did in fact provoke considerable muscular fatigue. Of course, it is still possible that discomfort originating in the passive structures added importantly to the subjective perception, but it is clear that the stress placed on the muscles was by no means irrelevant.

On the other hand, although the present study placed the emphasis on the strain borne by the muscles of the shoulder in postures with arm abduction, that does not necessarily mean that the loads placed on the passive structures may be disregarded. The existence of a sizeable strain acting on those passive structures was revealed by a highly relevant fact: it was at the insertion point of the medial deltoid muscles where most of the subjects reported to have experienced the sensation that forced them to give up the effort. This suggests that it was actually the tendons where the extreme discomfort was experienced, particularly at the end of the trials at 30° and at 60°. Thus, Milner's posture and the one used in the present study are not unlike to the degree of converting the comparison here undertaken into a trivial quest, although this could certainly be the conclusion drawn from an understandably misleading first impression.

8.2.1.2 Differences between the experimental designs

Another factor that was seen as a possible contributor to the discrepancy between the two studies was of methodological nature: an important difference between the experimental design applied by Milner and the one used in the present study. Milner (1985, p 130) used what he called a 'secondary task': the playing of a computer game. This was incorporated into the experiment with the overt aim of staving off boredom in the subjects whilst they remained in the required posture. However, in the view of this researcher, no matter how carefully Milner attempted to control for this factor, the use of the computer game introduced a strong element of added motivation on his subjects, such that ultimately he was not measuring the capacity of the subject to endure the postural load per se, but rather the desire to stay on playing the game. Since Milner's model was based entirely on the length of the posture holding capacity, it is obvious that any factor which contributes to get the subject to stay for longer will ultimately affect the size of the recovery.

This point was, in a rather unintended way, proved by Milner himself when he compared the endurance to the posture he tested when it was combined with the performance of a mechanically controlled tracking task and with the playing of a video game. Milner found that the subjects' endurance was obviously affected by the interest they accorded to the subsidiary task, and he wrote (Milner 1985, p 110): "The conclusion that one can draw from this is that task interest appears to directly influence Maximum Holding Time. Holding time in turn effects (sic) recovery". It was precisely because of this possibility that the decision was made by this researcher not to use any form of distracter (or subtle motivator), but to have the subjects endure the postural loads proper, and take care of the motivation by explaining clearly, and reinforcing as often as deemed necessary, that such was precisely the main aim of the investigation.

Nevertheless, this decision was not adopted in a vacuum, behind it lay the interest on carrying on with the research and attempting to prove the point. Unfortunately, time limitations impeded the completion of this goal, but this is clearly an open path through which the interest on static postural loading could be pursued further. However, the results obtained in the present investigation, in respect of the repeatability of the MHT (to be addressed in the next main section of this chapter) leave room to believe that the endurance to the postural loads will be fairly consistent despite (although it could as well be written 'because of') experimental conditions that offer no other incentive to remain in that posture than the sole desire of doing so.

8.2.1.3 The gender of the subjects

The gender of the subjects was indeed a major methodological difference between this study and Milner's. The ulterior aim behind the testing of Milner's model on female subjects was to then carry on and, assuming the model had proven its validity, to extend the present study into multiple cycles work/rest/work, until the development of exhaustion. However, seeing the unsoundness of the model, the researcher decided to turn round his priorities and look into the likely causes behind that result. But the probe into the possible differences related with the gender of the subjects was not abandoned altogether, since the main experiment involved male and female subjects, only this time looking for the influence of the gender factor on the consistency of the maximum holding time.

Nevertheless, there is no denying of the fact that replicating Milner's experiments on male subjects could produce some surprising results, and it is indeed unfortunate that there was not enough time to complete such a study. Nonetheless, this is a quite an obvious lead for further work on the matter of postural work and recovery.

8.2.2 Soundness of Milner's underlying assumptions about the model

The results obtained in the present study also raised two important issues about the foundations of the model proposed by Milner. The first one concerns the possibility that Milner's conceptions about the structure of the model were tinted by his being over-reliant on the apparent relationships between the length of the maximum holding time, and the lengths of initial holding and rest in the combinations he tested. Milner incorporated those relationships into the model, apparently taking for granted that for any two combinations with the same length of initial holding a larger recovery should follow the longer rest. However, this notion was roundly negated by the behaviour of subject No. 3 in the first experimental stage. As shown in chapter 5 (table 5.6), this person achieved exactly the same MHT on the three occasions it was measured. However, in the eight combinations of initial holding and rest she completed, her recovery always went against the expected trend, since she achieved the more recovery with the shorter rest (table 4.9, chapter 4). This then leaves wide open the possibility of Milner's model being based on rather shaky grounds.

Second issue: Milner (1985, pp 159-160) stated that the structure of the model was in agreement with the findings from other studies that have looked into the endurance to isometric exertion, and that the exponential term included in the equation was there to account for the effects of the physiological factors on the rate of recovery. Therefore, it is quite significant that the attempts at improving the fit between the model and the results of this study, by means of a slight adjustment to the constant in the exponential term, were unsuccessful. This approach was taken following the rationale expressed by Milner in the sense that the fundamental nature of the physiological changes that occurred during the exertion should not be very different from one posture to the other.

The final result of the intended modification was that even if the constant in the exponential term $e^{-0.164(HT/Rest)}$ was substituted by the value -1.108, meaning a reduction of two thirds of the term's power, the fit between the model and the data was extremely poor. Thus, the failure to achieve a good fit between the experimental results and the model the way it is structured (which is meant to predict those very results) raises the question of whether Milner was right in considering that the model could account for the physiological events occurring in instances of postural holding, irrespective of the posture being held.

8.2.3 Conclusions

The weight of the evidence produced by the experimental work carried out in the first stage of the present study leads to three conclusions:

i) Contrary to Milner's claims, the model for the prediction of recovery from purely postural work that he developed from and tested on a stooped posture, could not predict with a reasonable degree of accuracy the recovery to be achieved when the posture changed to an upright stance with both arms abducted at 60°.

ii) The underlying assumptions on which Milner based the model are not as sound as he purported them to be, particularly those referred to the relationship between the degree of recovery and the length of work and rest calculated in function of the maximum holding time.

iii) Milner's claim regarding the model as capable of accounting for the major physiological effects of purely postural work is most likely wrong.

8.3 CONSISTENCY OF THE ENDURANCE TO POSTURAL LOADING

8.3.1 Identification of the main influences on the endurance to postural loading

The trials performed during the second experimental stage of this investigation were designed to assess whether the maximum holding time (MHT) is a reliable and repeatable measure of endurance to purely postural loads. The work was also aimed at measuring the effects of the gender of the subject and the extent of the arm abduction on the length and consistency of MHT. The details of the methodology used to complete the experimental work were described in chapter 5.

The results showed that at the three levels of arm abduction studied, the male subjects were, on average, capable of enduring the effort for longer than the females. In this regard, however, there was a wide spread in the maximum holding time between individuals, so that some of the female subjects could endure the effort for much longer than some of the males.

The salient feature in the results of the holding trials to exhaustion was the variability of the maximum holding time. For the whole sample of ten subjects (5 male, 5 female) the coefficient of variation at 30° was 65.3%, at 60° it was 64.5% and at 90° it was 45%. These values also show that the variability of the response showed a reduction with the increase of the abduction angle. But in some cases the variability was evident not only in the comparison between subjects, but even within each individual. Thus, for the three measurements at 30° the coefficient of variation for individual subjects ranged between a low of 3% and a high of 17.7%; at 60° the range was between 1.2% and 18.7%, and at 90° it was between 5.4% and 26.3%.

Remarkably, despite such extent of variation between subjects, when the endurance capacity was considered for the overall sample, it showed to remain consistent throughout the repeated measurements. The analysis of variance on the length of MHT reached during each of the repetitions yielded F values of 0.18 ($p= 0.835$), for the tests at 30°, of 0.22 ($p= 0.807$) at 60°, and of 0.04 ($p= 0.965$) for the tests at 90°. This led to conclude that at group level the maximum holding time is highly repeatable and is therefore a reliable measure of the endurance to purely postural loads.

8.3.2 Comparability of the results with those from other studies

The coefficients of variation for the maximum holding times observed in this investigation are not very different from those found in other studies. For example, the endurance times of six female subjects to abduction at 90° reported by Hagberg (1981a) had a coefficient of variation of 0.439, which is slightly higher than the 0.308 affecting the endurance to the same abduction angle by the female subjects in this study. Fallentin and Jørgensen (1992) studied the endurance of male subjects to elbow flexion and extension at strengths of 10% MVC and 40% MVC. They found coefficients of variation of 0.504 and 0.414 for the flexion and extension at 10% MVC, respectively. At 40% MVC the coefficient of variation was 0.304, for the endurance to both manoeuvres. Those values, again, are not very different from the coefficient of variation for the data collected from the male subjects in this study: 0.535 for the abduction at 30°, 0.486 at 60° and 0.418 at 90°. Besides, as in the present study, Fallentin and Jørgensen (1992) also found that the increase of the load had the effect of reducing noticeably the dispersion of the endurance to the effort. Bearing in mind that the exertion they studied involved the continuous application by the subject of a constant force, then it is clear that the variability between subjects is significantly reduced by an increase of the level of exertion, be it purely passive as in the present study, or active as in Fallentin and Jørgensen's.

Regarding the endurance time itself, it is difficult to assess how typical the results obtained in this investigation are, since the number of other studies dealing with the endurance to purely postural loads imposed on the upper limb is very limited. Chaffin (1973) studied a group of 5 young males, asking them to hold both arms abducted at angles of 30, 60, 90 and 120°, until reaching what Chaffin defined as 'class II muscle fatigue', a state of discomfort described as "cramping continuous with deep hot pains intermittent". The average times in which the subjects felt to have reached that stage are shown in table 8.1, together with the corresponding average of the maximum endurance times of the male subjects in the present study.

Table 8.1 Average length of time (minutes) for which 5 male subjects could hold both arms abducted before discomfort forced them to stop.

	Arms abducted at		
Study	30°	60°	90°
Chaffin (1973)	68	25	10
Serratos (1994)	36	22	9

It is remarkable that whilst the average holding times at 60° and 90° were practically the same in both studies, at 30° the subjects in Chaffin's study held it for almost double than the subjects in the present study. However, it is difficult to ascertain what could be the cause for such disparity. There was certainly an important difference between the postures adopted by the subjects. Whilst in Chaffin's study they placed their forearms close to the body, with the hands almost touching their chest, in this study the forearms were extended in the saggital plane, putting the hands at forearms' reach. This meant that in Chaffin's posture the centre of gravity of the forearms was kept closer to the body, and this probably reduced the moments acting on the glenohumeral joint. However, if such biomechanical advantage did exist, that would only add to the puzzle, for whilst it could explain the difference in the endurance at 30°, it would certainly raise the question of why its influence was so powerful at the lowest abduction angle, and then practically disappeared as the angle increased.

Obviously, the difference in endurance at 30° has a large effect on the shape of the relationship between the abduction angle and the endurance time. This is illustrated in figure 8.1 where the average endurance reported by Chaffin (1973) is plotted alongside the average MHT for the males in the present study. The results obtained by Chaffin (1973) suggest that the endurance decreases in exponential fashion when the abduction angle increases (fig 8.1 a); however, the results of this study show that the endurance decreases linearly with the increase of the abduction angle (fig 8.1 b). Fitting a linear regression model to the data obtained during this investigation yielded the equation MHT = 49.8 - 13.7[ANGLE], with R^2 of 44.1%. Although this value is indicative of a fairly low explanatory power, it is actually quite acceptable, bearing in mind that the major source of variance in the raw data is the variability between subjects.

Hagberg (1981a) asked seven female subjects to hold to their limit of endurance the right arm abducted at 90°, with the elbow flexed at 90°, and the forearm in a vertical position and rotated internally. He reported holding times ranging from 8.4 minutes up to longer than 60 minutes, with an average of 17.3 minutes. These values contrast sharply with those observed in the present study, where the average endurance of the female subjects at 90° ranged from 3.1 minutes to 6.2 minutes.

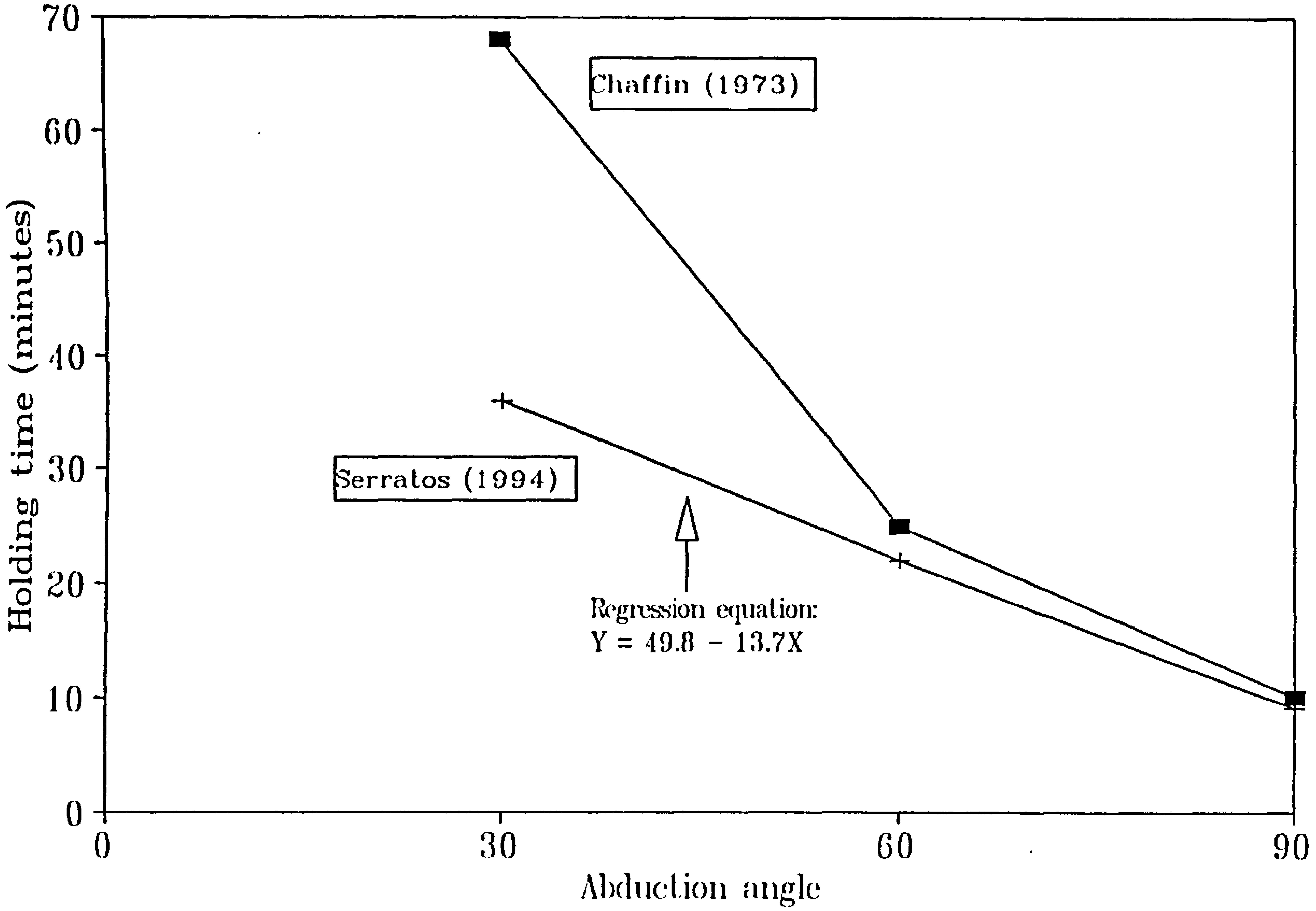


Figure 8.1 Plot of the average limit holding time for abduction of both arms measured on samples of 5 young males

There is a closer agreement between the endurance times found in the present study and those reported by Hansson et al (1992). They studied three groups of female subjects (11 subjects in each group), who were asked to raise their arms by their sides until they were horizontal and hold them in that position, keeping their elbows and wrists straight. Hansson et al did not report the actual endurance achieved by their subjects, only the median value for each of the three groups, which were 355, 411 and 457 seconds. These are not very different from 365 seconds, the median value of the endurance achieved by the female subjects in this study when they held the posture at 90°.

Corlett and Manenica (1980) reported on the maximum length of time that female subjects could spend performing a tapping task whilst adopting a series of postures, one of which was very similar to the posture with abduction at 90° used in the present study. Concretely, Corlett and Manenica pre-defined their postures of interest in function of the height at which the hands would be located (measured as a percentage of the subject's height) and of their distance from the body, expressed as a percentage of the arm reach, location that was to be achieved through a combined abduction and extension of the shoulder. If the postures investigated in the present study are expressed in this way, the abduction at 90° becomes one with hand height of 98% of shoulder height and hand distance equal to around 60% of arm's reach, and this is very close to a posture with hands located at 100% shoulder height and 75% arm reach as defined by Corlett and Manenica. They reported that the maximum holding time for that particular posture was around six minutes, and this tallies well with the average MHT of 5.8 minutes achieved by the female subjects in abduction at 90°. One further agreement between the two studies is that the subjects holding the posture devised by Corlett and Manenica reported the presence of unbearable discomfort in the shoulder area, which was precisely the effect wanted from the postures used in the present study.

The similarity of the results from the present study with those reported by Hansson et al (1992) and by Corlett and Manenica (1980) makes their difference with the results of Hagberg (1981a) even more difficult to interpret. However, a possible explanation for such a large difference in maximum holding times may be found by comparing the postures involved in the two investigations. Hagberg did not state whether the abduction he studied occurred in the coronal or in the scapular plane, but the diagrams he presented suggest it was in the latter. If that was the case, then his subjects held important biomechanical advantages over those involved in this study (and, incidentally, those in Hansson et al's) for the abduction in the scapular plane keeps the glenohumeral joint in the neutral position, avoiding the impingement of the greater tuberosity against the acromion (Perry, 1988), a circumstance that leads to painful sensations rather quickly. Besides, during abduction in the scapular plane the three portions of the deltoid muscle pull in the same direction, as opposed to the abduction in the coronal plane when their lines of action are not convergent (Perry, 1988). This creates a greater load on the medial portion, a factor that in turn accelerates the appearance of fatigue.

Thus, the evidence provided by the comparison against this handful of studies shows that the maximum holding times measured in the present study fell very much within the range that might be expected. Incidentally, this finding provides further proof of the reliability of the maximum holding time as a measure of the endurance to postural loading of the shoulder region.

8.3.3 Factors that might explain the variation in the maximum holding time

The search for a model that could explain the variations of the maximum holding time met with a good deal of success: a linear regression model was found that could explain nearly 60% of the variations for the overall sample, and the explanatory power rose to 80% when the sample was split into the two gender sub-samples. In the three cases the explanatory power was provided mainly by three factors: the abduction angle, the subject's height and the height to the shoulder. However, since the last two variables exhibited an extremely strong correlation (indeed, for the vast majority of people their shoulder height is a function of how tall they are), practically all the explanatory power stemmed from the abduction angle and the subject's height.

It is convenient to emphasise that the search for such model was above all an effort to identify the factors most likely to explain the variations observed in the endurance to postural loads, and was not intended as a regression model that could be used as a predictive tool. The main reason to be wary in this respect is because of the prominent role accorded to the stature of the subject, which could be wrongly interpreted as meaning that the taller the person is, the more resilient he or she should be to the effects of postural load. However, this relationship would hold mainly for the female subjects: when the correlation coefficient between stature and MHT was calculated separately by gender, the value for the males was 0.118 whilst for the females it was a much stronger 0.466. However, such a moderately strong positive correlation between stature and endurance capacity was not found among the female subjects who participated in the first series of trials of this study, where the correlation coefficient equalled only 0.048.

Milner (1985) also reported conflicting evidence about the relationship between stature and endurance capacity. He found a correlation coefficient as high as -0.891 for a group of nine subjects, but it decreased to -0.264 for another group of six subjects. Besides, the relationship he observed went opposite to the one seen in this study: the taller the subject was, the less they seemed capable of coping with the postural loads. This last effect might have been linked to the posture Milner studied, since it could be assumed that the taller the person is, the larger the proportion of the body mass represented by the upper part of the body, and this will impose larger moments on the hip and low back regions when the person bends forwards to the extent required by Milner's posture. However, he also found that there was no significant correlation between the subject's weight and the length of the MHT, as neither was in the present study.

In summary, the relationship between stature and endurance to purely postural loads appears to be a feature that will change not only from posture to posture, but even when measured for different groups of subjects in the same posture. Thus, the evidence currently available makes it inadvisable to attempt the prediction of MHT for postural effort on the basis of its relationships with the anthropometric features.

8.3.4 Conclusions

Three conclusions may be drawn from the foregone discussion:

1) As may be expected from what is essentially an individual trait, the MHT exhibited a wide variability. This was evident not only in the comparison between subjects, but also in considering the results from the repeated measurements on some of the subjects;

2) A large proportion of the observed variation was explained by a combination of the change in the abduction angle and the height of the subject. However, the extent of the variation that could be explained by the anthropometric factors considered in this study was very different between genders.

3) Nevertheless, when it was considered as a feature for a group of subjects rather than for each individual, the maximal holding time emerged as a repeatable measure.

8.4 THE RELATIONSHIP BETWEEN POSTURAL ENDURANCE AND THE PERCEIVED GROWTH OF DISCOMFORT

8.4.1 Nature of the relationship between discomfort ratings and holding time.

The results from the main experiment (presented in chapter 6) showed that when a person holds a static posture which places the shoulder region under stress, there is a highly significant linear relationship between the discomfort ratings returned by the subject - in this particular case reflecting the growth of the discomfort affecting the medial deltoid muscle- and the length of time spent in that posture.

A linear relationship between discomfort (or pain) ratings and the passage of time in different forms of exertion has been found in other studies, conducted in a variety of settings. For example, Caldwell and Smith (1960) studied the subjective perception of pain during handgrip sustained to exhaustion, and using a 5-point scale found that the ratings of pain increased linearly with the time. Kirk and Sadoyama (1973) studied two forms of static forceful exertion to exhaustion, which was applied with either one arm or both arms, engaging two very different muscular groups. They also used a rating scale with only five marks on it, finding that at low levels of strength there was a linear relationship between discomfort ratings and the passage of time.

Both Barbonis (1979) and Milner (1985) reported the existence of a linear relationship between discomfort ratings and the holding time. In their respective studies of muscular loading during long-term stooping, they collected information on the time course of discomfort using the 6-point scale for discomfort/pain rating devised by Corlett and Bishop (1976). The structure of that scale is shown in figure 8.2. It is noteworthy how it very much looks like a half-sized version of the 10-point scale later developed by Borg in 1982, which was used in this study.

0 = No Discomfort/Pain

1 = Just Perceptible Discomfort/Pain

2

3 = Moderate Discomfort/Pain

4

5 = Intolerable Discomfort/Pain

Figure 8.2 The 6-point scale for discomfort rating devised by Corlett and Bishop (1976)

Harms-Ringdahl et al (1983) studied the increase of discomfort and pain in the passive structures of the joints, as perceived by subjects who had the joint of either one elbow or one knee moderately loaded whilst kept in an extreme position. They used Borg's 10-point scale, only they had to restrict the subjects to go no further than a rating of 7, to avoid the risk of permanent tissular damage. Harms-Ringdahl found that the discomfort grew following a straight linear pattern for both joints tested; the coefficient of correlation for the data generated from the elbow was $r= 0.9981$, and for those from the knee it was $r=0.9978$.

Manenica (1986) described the treatment of the data gathered during the study in which female subjects held one of seven postures to the limit of endurance whilst performing a tapping task. Every 30 seconds the subjects returned ratings of discomfort using the 20-point scale proposed by Borg (1973) and, at the end of the experiment, were asked to identify on a body map the body region worst affected by discomfort. The analysis of the discomfort ratings so collected showed that the discomfort grew linearly during the holding of the postures. The overall regression line was of the form $y= 0.13 + 0.199x$, with a correlation coefficient of 0.99 ($p< 0.001$). The equation of the regression line fitted to the discomfort ratings collected during the main experiment (second stage) of the present investigation was $y= -0.5 + 0.107x$, with a correlation coefficient $r= 0.96$. Allowing for the fact that Manenica (1986) used a 20-point scale for the rating of discomfort, and in this study a 10-point scale was applied, the similarity between the two regression equations is quite remarkable.

Thus, ample evidence has been produced of the fact that the degree of discomfort (or pain) generated by the performance of static exertion is perceived by the subject to grow in a linear fashion. This pattern of linear growth has been demonstrated in practically any form of isometric exertion, be it a situation where the muscular stress derives from the load created by the holding of a posture (Barbonis, 1979; Milner, 1985; Manenica, 1986), the application of constant force by a well defined group of muscles (Caldwell and Smith, 1966), the use of force to counteract an external load (Kirk and Sadoyama, 1973) and even the passive supporting of loads (Harms-Ringdahl et al, 1983).

However, such uniformity of subjective perception was questioned by Kinsman and Weiser (1976), who attributed it to the methodology applied to collect the ratings of the sensation involved. Nevertheless, this issue has been addressed (Menzer et al, 1969) and it appears that the linear increase of the subjective perception is a true and valid perceptual phenomenon. Furthermore, this impression is backed by the fact that the linear relationship has been demonstrated in a wide variety of experimental settings, fitting regression models with very high explanatory power, such as the 99.6 % in Harms-Ringdahl et al (1983), 98% in Manenica (1986), and 92% in the present study.

8.4.2 Use of the relationship between endurance capacity and subjective perception as a modelling tool

The shape of the relationship between discomfort ratings and percentage holding time found in this study was represented by the regression equation

$$\text{Discomfort Rating} = -0.509 + 0.107\ [\%\ \text{MHT}],$$

which has a standard error of the estimate (S_{ee}) equal to 1.018 and a coefficient of determination (R^2) of 91.5%. The regression line for this equation is shown in figure 6.4. Forcing the model through the origin changed the equation into

$$\text{Discomfort Rating} = 0.0994\ [\%\ \text{MHT}];$$

S_{ee} changed to 1.052 and R^2 increased to 96.9%. The new regression line may be seen in figure 6.5.

This sort of equation has been shown as the true representation of the perceptual phenomena elicited by isometric muscle exertion (see section 8.4.1). Therefore, putting them forwards as a model for the perception of the development of fatigue in work situations with large postural demands makes a sensible proposition, not least because, following the trend pointed out in Corlett and Manenica (1980), those conditions are nowadays becoming more the norm than the exception in a wide variety of work settings (for examples, see section 1.6).

Thus, a model based on either of the regression equations given above would say that, on average, it might be expected that the usage of every 10% of one person's capacity to endure the postural loading of the shoulder would reflect as the increase of one unit in their ratings of perceived discomfort, measured on the 10-point scale used in the present study. However, a more useful form for the model would be one that allows to calculate what proportion of the maximum endurance to the posture has already been used, based on the discomfort ratings returned by the person while working in that posture. This may be achieved by simply reversing the role of the variables when calculating the regression model. When this procedure was performed on the data pool gathered during the main experiment, the resulting regression equations, and their corresponding statistical parameters were:

$$\% \text{ MHT} = 8.70 + 8.58 \text{ [Discomfort Rating]}, \ S_{ee} = 9.1, \ R^2 = 91.4\%$$

for the regression model fitted with intercept to the vertical axis, and

$$\% \text{ MHT} = 9.76 \text{ [Discomfort Rating]}, \ S_{ee} = 10.4, \ R^2 = 96.9\%$$

when it went right through the origin.

Obviously, neither the make-up of the relationship between the variables that is expressed by the equation, or the value of the coefficient of determination have changed; however, the structure of the equation with free intercept to the Y-axis deserves some comment. The value of 8.70 for such intercept implies that, on average, nearly 9% of the endurance capacity has already been used when the subject returns a discomfort rating of zero, that is when he or she is actually feeling no discomfort at all. At the other end of the scale, by contrast, the highest value of discomfort rating will be assigned when over 5% of average endurance capacity still remains. This apparent nonsensical situations derive from the data used to calculate the regression models.

On the one hand, they include a number of data pairs in which the discomfort rating is zero even though an appreciable proportion of the holding time has elapsed, which happened mainly during the trials performed mainly at 30° - but in some cases also at 60°- by subjects with long maximum holding times. On the other hand, and also mainly during those long-duration trials, the subjects reached the upper end of the scale of discomfort ratings but still found they could endure the demand posed by the posture before the sudden appearance of the maximal discomfort.

These two awkward situations are avoided by giving the model the form of the equation forced through the origin. Besides, doing so will reflect more accurately what actually happened during the main experiment, in that in 85 trials out of the 89 performed the subject reported total absence of discomfort at the start of the holding. Also, the model will predict that (on average) when the person has used 100% of the endurance capacity they will return a discomfort rating of 9.76, which in reality would be a 10 on the scale. However, before stating overtly that a model with general applicability has been found, it is necessary to address the issue of the effect that the data collected during the longer trials at low abduction angles appeared to exert on the outcome of the regression model. This is easily done by fitting a separate regression model to the data collected at the three abduction angles and comparing their structure to that of the overall sample.

The regression equation for the separate sub-samples and their statistic parameters were as follows:

i) Sub-sample of data collected at 30°

$\% \text{MHT}_{30} = 9.86$ [Discomfort Rating]$_{30}$; $R^2 = 96.7\%$; $S_{ee} = 10.67$; $F= 1.02$.

ii) Sub-sample of data collected at 60°

$\% \text{MHT}_{60} = 9.55$ [Discomfort Rating]$_{60}$; $R^2 = 97.0\%$; $S_{ee} = 10.25$; $F= 0.98$

iii) Sub-sample of data collected at 90°

$\% \text{MHT}_{90} = 9.90$ [Discomfort Rating]$_{90}$; $R^2 = 96.5\%$; $S_{ee} = 11.67$; $F= 1.11$.

The corresponding information for the overall sample is:

$\% \text{MHT}_{\text{ALL}} = 9.76$ [Discomfort Rating]$_{\text{ALL}}$; $R^2 = 96.9\%$; $S_{ee} = 10.42$

There is no obvious difference between the three equations for the sub-samples and that for the overall sample. They all lead to practically the same maximum value for the 'dependent' variable, the values of R^2 are very similar, as are the values of See. Although for the sample at 90° the value of F (calculated as the ratio of the squared values of S_{ee}) seems suggestive of a significant difference, it has to be considered that this was the sample with the smallest number of data and these lay scattered over a very wide range, and this had a disproportionate impact on the value of the standard error of the estimate. However, neither the value of the slope coefficient or that of R^2 for this equation were actually very different from the corresponding coefficients for the other two. Therefore, the best means available to model the relationship between the subjective perception of the increase of discomfort during the performance of a purely postural effort is the equation

% MHT = 9.76 [Discomfort Rating], which is presented graphically in figure 8.3.

To put the model to use, once the MHT for a posture is known, it must be decided which is the highest level of discomfort that will be allowed and limit the duration of work to the corresponding proportion of MHT. For example, if the discomfort build-up is to be kept from going any further than 'moderate', which is the description attached to a rating of 3 on the scale, then the postural work should be no longer than 30% of MHT.

Alternatively, the model might be used in the way proposed by Manenica (1986), as a predictor of maximum endurance to postural load. To do this, all is needed is to measure how long it takes for the person to judge that the discomfort rating has increased by one unit, that time would then represent a 10% of the maximum endurance, and multiplied by 10 should tell how long that would be. Once this is known, measures could be taken to arrange the working conditions in a way that ensured that the person should not remain in a fixed posture for longer than a certain proportion of their maximum endurance to it.

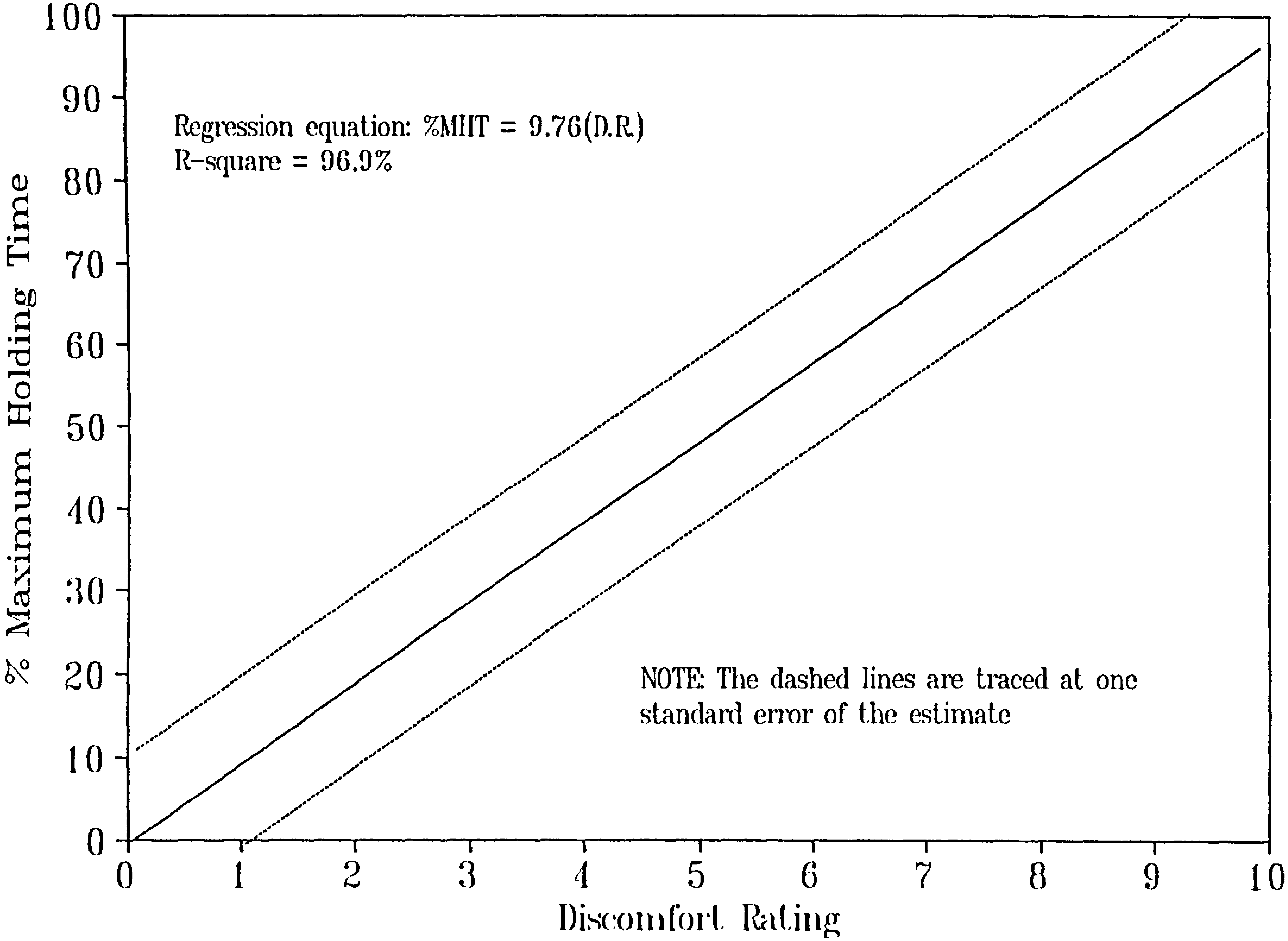


Figure 8.3 Graphical representation of a model that predicts the proportion of endurance capacity for postural work used/remaining in function of the discomfort rating on Borg's 10-point scale.

Naturally, it is unrealistic to think that either mode of use could be implemented on an individual basis. This would require that either the MHT (first approach) or the interval for the unitary increase of discomfort rating (second approach) be measured for each and every person engaged in the work to which the model is applied, then keeping track of their working time, making sure to call them off work once they have reached the pre-set time target, give them the length of rest necessary until the discomfort has disappeared, and then put them back to work, so starting the cycle again. Therefore, the model will have to be implemented based on the maximum holding time averaged over a representative sample of the people among whom a problem of excessive postural discomfort is most likely to develop, using this information to establish guidelines for the length of working time in such a way that the largest proportion of workers are included within them. This is precisely the approach suggested in Dul et al (1993).

8.4.3 Frequency of appearance of maximum discomfort in the muscles studied

During both series of experimental sessions which constituted the present investigation, at the end of each trial the subject was asked to mark on a copy of the body mapping (fig 3. 6) the area where they had perceived the discomfort to be unbearable. That area was then equated by the researcher with the superficial muscle or muscles it comprised. Table 8.2 presents a summary of the information provided by the 8 female subjects who participated in the first experiment and the 10 subjects (5 male, 5 female) who took part in the second one.

Table 8.2 Details of the superficial muscles most affected by discomfort in the two experimental stages of the study.

Number of reports of maximal discomfort on:		
	First Experiment (64 trials)	Second Experiment (90 trials)
Right medial deltoid	50	82
Left medial deltoid	38	74
Right posterior deltoid	14	6
Left posterior deltoid	9	6
Right trapezius	20	0
Left trapezius	12	0

The information presented in table 8.2 shows that there was a basic agreement between the results of the two series of holding trials with different groups of subjects, since in both cases the medial deltoid muscle was identified as the site of maximal discomfort more times than any other muscle.

That maximum discomfort affected the medial deltoid more often than any other of the superficial muscles located in the shoulder/nech area may be explained from a knowledge of the functional anatomy of the muscles of the shoulder region. The medial portion of the deltoid muscle is the prime mover in all modes of arm elevation, with assistance from both the anterior and the posterior portions (Perry, 1988; Kronberg et al, 1990). Abduction in the coronal plane activates all three portions and, when it is below 90° the posterior portion plays a more important role than the anterior portion, and when the arm goes higher than 90° that relationship is reversed. Since the trials involved abduction between 30° and 90°, this explains why, of the three portions of the deltoid, it was the medial portion which was most often mentioned as the site of maximal discomfort, followed by the posterior portion, and the anterior portion was not mentioned at all by either of the two groups of subjects.

To explain the mention of the trapezius muscle as a site of maximal discomfort, it has to be considered that arm abduction is accompanied by outward rotation of the scapula, and this adjustment is accomplished by the action of the upper portion of the trapezius (Lucas, 1973). Besides, although this muscle is activated throughout the whole range of abduction movement, its action becomes more marked when the angle is higher than 30° (Perry, 1988).

The reference to the participation of the trapezius muscle in arm abduction highlights an important feature present in table 8.2, which is well worth considering. It is remarkable that, whilst the subjects who took part in the first series of trials pointed at the trapezius muscle as a site of extreme discomfort (right arm on 20 occasions, left arm on 12), those involved in the second series of experiments did not produce a single report of the kind. The reason for this difference could lie in the basic design of the two experiments. The first experiment involved two holdings in succession, the first of them shorter than, or at most equal to the subject's maximum holding time (which was known from an earlier measurement), with the second holding invariably leading to exhaustion. Those two holdings were separated by a period of rest that, at the most, could be as long as the first holding time. On the other hand, the second series of experiments involved a single holding, always lasting to the endurance limit of the subject. It is therefore quite likely that the higher incidence of maximum discomfort on the trapezius muscle reported during the first experiment was due to the cumulative effects of the two holdings on the trapezius. The proposed mechanism is that, although the discomfort imposed on the trapezius muscle by the first holding was not of a magnitude as to make it feel equal to that in the deltoid muscle, the addition of the second holding (to exhaustion) brought it to the same level. At the moment this stands only as a tentative explanation, but the results from the study of the myoelectrical activity carried out during the second series of experiments appears to lend it a good deal of credence. The foundation for this assertion may be found in the graphs shown in Appendix D , which show that in many trials the largest values of RMS amplitude corresponded to the trapezius muscles. Assuming that the degree of muscular activation is measured by the value of RMS amplitude (Basmajian and De Luca, 1985), then it is true that the trapezius muscle was the most heavily activated during the trials, and should the experiment be extended to measure EMG changes during at least one single combination work/rest/work, then the values of RMS amplitude for this muscle would be even larger than for the two deltoid muscles. Indeed, this looks quite an attractive proposition for further work on the issue of the measurement and evaluation of pure postural loads.

8.4.4 Conclusions

Three basic conclusions may be established regarding the relationship between the passing of the holding time, the concomitant increase of the subjective perception of discomfort during it, and the parts of the body where this was most heavily experienced. Those conclusions are:

i) There is a highly significant linear relationship between the passage of the holding time and the growth of discomfort as reported by the subjects;

ii) So strong is that relationship, that it may be advanced as a tool for the prediction of the length of time it would take for a person working in a given posture to reach a certain degree of discomfort. Alternatively, it could be used to predict the maximum length of a postural effort, based on the time it takes for the discomfort ratings to increase by one unit;

iii) During the holding to exhaustion of a posture with abduction of both arms, the maximum discomfort affects mainly the medial portion of deltoids muscle and the descending part of the trapezius muscle. This is a result in complete accordance with the anatomical features of the glenohumeral region.

8.5 ELECTROMYOGRAPHIC MANIFESTATIONS OF MUSCULAR FATIGUE

8.5.1 The relationship between EMG changes and the holding time

In this study, the changes in the mean power frequency (MPF) and in the RMS amplitude of the EMG signals collected from the descending part of the trapezius, the medial and the posterior deltoid muscles, in both arms were used to assess the presence of fatigue as a result of the holding to exhaustion of a standing posture, with both arms abducted. Trials were conducted on male and female subjects, in three well differentiated conditions, defined by the angle at which the arms were kept. In most of the 89 trials successfully completed, for all the six muscles studied, there was a significant shift of MPF towards lower values and/or a significant increase in the RMS amplitude, changes that are widely accepted as signs of the existence of localised muscular fatigue (Basmajian and De Luca, 1985).

However, it was not possible to find a model that adequately describes the changes that were observed in the EMG signal over the holding time in the course of the 89 trials. Although, in many particular instances, the change in the EMG variables over time exhibited a strongly linear pattern, when the results from all the experiments were pooled together the linearity was no longer evident.

The difficulty in fitting a model to describe the relationship between changes in the EMG signal and the passage of time appears to be a feature common to several studies involving exertion with muscles of the shoulder. Thus, Gerdle et al (1988) monitored the EMG responses from the trapezius, anterior deltoid, and infraspinatus muscles during the isometric forward flexion of the right arm at increasing angles. They concluded that, at low output levels, there is not a definite pattern to the relationship between the changes in the EMG signal and the passage of time. Hansson et al (1992) studied the EMG responses from the trapezius and medial deltoid muscles during the abduction of both arms at 90° sustained to the limit, and they too could not find a relationship between the endurance time and the EMG parameters. Takala et al (1993) studied the behaviour of the EMG signals during a test of maximal endurance to arm flexion with a weight attached to the wrist, collecting signals from the descending and the lower parts of the trapezius muscle, from the infraspinatus and from the anterior deltoid, all on the right arm. Their attempts at fitting a variety of regression models to the relationship between endurance time and EMG changes were also unsuccessful.

Nevertheless, there are also reports of strong linear correlation between the EMG parameters and the passage of time, although these come mainly from studies involving the exertion of constant force with the muscles of the arm and forearm. Jørgensen et al (1988) mention the results of long-lasting, low-level contractions performed on a dynamometer, with collection of EMG signal from the right biceps and triceps muscles. They found that the changes of both MPF and RMS were linear. Hasson et al (1989) studied the changes in the EMG signals during handgrip contractions applying a force equal to 50% MVC on a dynamometer; the typical endurance time of their subjects was about 90 seconds. They collected EMG signals from the right flexor digitorum superficialis muscle and found a near-perfect linear correlation between both MPF and RMS amplitude and the endurance time. Caffier et al (1993) monitored the EMG signals from the right biceps muscle during sustained static contractions at 4%, 8% and 15% MVC when the target duration of the exertion was one hour. At all the experimental conditions the relationship between RMS amplitude and time was nearly linear, but it definitely was non-linear between MPF and time.

This selection of references confirms what De Luca asserted back in 1985 (De Luca, 1985, p 270): the relationship between the EMG parameters and the exertion time does not appear to conform to a single model, although it may indeed conform to several different patterns, mostly depending on the muscle studied and the mode of the exertion.

Thus, the results from the present investigation add to the evidence that points to the extreme difficulty in finding an unitary model which may describe the changes that occur in the electromyographic activity of the muscles of the shoulder region during isometric exertion involving low levels of strength.

8.5.2 The extent of the changes in the EMG signal

The changes in the EMG signal observed during the majority of the trials in this study were significant enough to demonstrate that purely postural exertion does in fact provoke muscular fatigue. The average increase of RMS amplitude ranged from 22% to 68%, and the average decrease of the mean power frequency ranged from 8% to 16%. However, these changes were, in general, smaller than those reported from similar studies, as will be shown in the next two sections.

8.5.2.1 Size of the changes in RMS amplitude

Hagberg (1981a) studied the changes in RMS amplitude of the EMG signals during abduction of the right arm at 90° held to exhaustion (no force was exerted by the subject). He collected signals from, among other muscles, the descending part of right trapezius and the right medial deltoid, calculating the ratio between the average values of RMS amplitude at the end of the exertion and at the beginning of it, and found an average ratio of 2.3 for the trapezius muscle and 2.2 for the medial deltoid. In other words, the RMS amplitude had increased by 130% and 120% respectively. In contrast, the larger increases in the absolute value of RMS amplitude observed in the present study (table 7.9) occurred in the left medial deltoid (44%) and in the left trapezius (68%). Hansson et al (1992) also studied the endurance to the abduction of both arms at 90°, with no force exerted by the subject. They found that the RMS amplitude of the trapezius muscle increased by as much as 150% of its original value, which is nearly two and a half times the larger of the increases seen in the present study. However, the changes of RMS amplitude in the medial deltoid muscles were not very dissimilar: 44% in this study and 50% in Hansson et al's. Indeed, this is just the same size as the change for the left posterior deltoid found in this study.

However, there is a more marked contrast with the changes that have been reported to occur during forceful muscular exertion. For example, Gerdle et al (1988) studied the EMG changes in the course of isometric shoulder flexion (10 seconds long), with the force applied by the subject increasing gradually up to 100% MVC. However, they made no direct reference to the magnitude of these changes, but from the graphs they presented it may be inferred that the increase in RMS amplitude for the trapezius muscle was as much as 400%, and around 100% for the anterior deltoid muscle. Jørgensen et al (1988) presented results from several studies of isometric exertions which involved the application of forces between 10% and 40% MVC. They mentioned increases in RMS amplitude of 150% for the right triceps muscle and of around 75% for the right biceps. Hasson et al (1989) studied the EMG changes provoked by sustained handgrip at 50% MVC, and found an increase of approximately 100% in the RMS amplitude from the flexor digitorum superficialis. Krogh-Lund (1993) studied the static flexion of the right elbow to exhaustion at two levels of strength, 40% MVC and 10% MVC, and collected EMG signals from the brachioradialis and the biceps brachii muscles. He reported increases of up to 400% in the RMS amplitude for the exertion at

10% MVC and, for the exertion at 40% MVC, the increases were in the range from 150% to 200%.

Lee (1987) took an approach to the issue of muscular fatigue which very much resembles the one applied in the present investigation. Rather than measure the fatigue provoked by the continuous and forceful isometric exertion of a single muscle group, he looked into the changes that occurred whilst the subjects worked at a microscopy station, in conditions that ensured they remained in a fixed posture for most of the time. The subjects worked continuously for 4 hours without a break, and EMG signals were collected from the descending part of both right and left trapezius muscles. By the end of that period, the RMS amplitude in both muscles showed an increase of 65% compared to the value it had before the work started, an increase practically equal to the average of 68% found in this study for the left trapezius, but still higher than the 39% observed in the right trapezius.

8.5.2.2 Size of the changes in Mean Power Frequency

Changes in the frequency parameters (the most frequently used being mean power frequency and median frequency) of the EMG signal are typically of smaller magnitude than those observed in the RMS amplitude, and not always reported as percentages of the initial value. Thus, Jørgensen et al (1988) reported a marked decrease in MPF for the triceps, from 96.1 to 69.9 Hz whilst the value for the biceps the MPF remained practically unchanged, going from

77.3 to 75.4 Hz. Gerdle et al (1988) produced graphic evidence of decreases in MPF that ranged from 4 Hz for the biceps brachii to 12 Hz for the anterior deltoid. Caffier et al (1993) reported that the static contraction of the biceps provoked decreases of MPF from 86.5 Hz to 80.1 Hz when the force applied was 15% of the MVC, and from 82.9 Hz to 79.1 Hz when the force was 8% of MVC. Hasson et al (1989) found that MPF from the right flexor digitorum superficialis decreased to nearly 50% of its original value. Hansson et al (1992) reported that the MPF from the medial deltoid decreased by approximately 70% of its initial value, whilst that of the trapezius remained almost unchanged. Krogh-Lund (1993) found that the exertion at 10% MVC provoked a 20% reduction in the median frequency, and that at 40% MVC it was reduced by between 55% and 80%.

Takala et al (1993) found that the median frequency of the signal from the descending trapezius decreased by 20% for their female subjects and by 29% for the males; in the anterior deltoid the decreases were of 36% for the females and 48% for the males. These findings are particularly relevant to those made in the present study, since their sample was also composed of male and female subjects. They observed that the changes were consistently larger for the male subjects than they were for the females, and the difference was statistically significant for the lower part of the trapezius and for the infraspinatus. In the present investigation, the changes in all the muscles studied were also significantly larger for the males than they were for the females, although this happened for the changes in RMS amplitude and not for those in the mean power frequency. Takala et al (1993) asserted that the reason for the changes being larger for the male subjects was not directly related to the EMG signals, but rather reflected the fact that the female subjects displayed a lower endurance to other physiological phenomena, such as the pressure on the pain-sensitive structures of the muscular system.

8.5.3 Likely causes for the change in the EMG parameters

When describing the results of this study (chapter 7), attention was drawn to the apparently different ability of the two EMG criteria applied to detect the onset of fatigue in the different shoulder muscles, that is, whilst in the trapezius muscle the increase in RMS amplitude reached significance on more occasions than did the decrease in MPF, the situation was reversed in the deltoid muscle.

A similar finding was reported by Hansson et al (1992) who, it is worth mentioning, carried out a study where EMG signals were collected from the descending part of trapezius and from medial deltoid muscles while female subjects held to exhaustion a posture that, whilst resembling the one used in the present study, was also different to it in an important aspect. The similarity between the two postures consisted in the requirement for the subjects to keep their arms abducted at 90° in the coronal plane, but the postures differed in the disposition of the forearm and hand, for whilst in Hansson et al's the elbow was fully extended and the hand hung freely, in this study's posture the elbow was flexed at 90° and the hand was aligned with it and exerting a very light pinch between thumb and forefinger.

To explain the predominance of RMS amplitude in the response of the trapezius muscle, Hansson et al (1992) referred to the role that this muscle plays in arm abduction, since besides participating in the abduction proper, the muscle also acts to adjust the scapula. In consequence, to stabilise the shoulder joint during the sustained abduction, the trapezius recruits new motor units to cope with the increased strain, even though the net joint moment remains constant. The newly recruited motor units may not be fatigued, may be of a different type or size, and/or may be at other distances from the electrodes. Since MPF depends on the average conduction velocity of the muscle fibres (Lindström et al, 1970; Sadoyama et al, 1983), the addition of new fibres obscures the changes in the conduction velocity of those fibres which have been active from the beginning of the exertion. In consequence, the signs of fatigue in the trapezius will be more obvious in the change of RMS amplitude, which depends mainly on the number and size of active muscle fibres.

The explanation offered by Hansson et al (1992) for the predominance of the MPF response in the deltoid muscle follows directly from the last assertion stated above: since the deltoid muscle is the main mover in arm abduction (Perry, 1988), it is heavily activated from the start of the abduction and has to recruit large numbers of muscle fibres, which remain practically constant whilst the arm is kept in a fixed position. Therefore, the deltoid muscle has no resource to additional motor units, and the reduction in the conduction velocity of the muscle units already activated will be reflected as a reduction in the frequency parameters of the EMG signal. This explanation for the difference in predominance may in fact fit with the findings from the present study, since it was observed that the majority of the subjects used a strategy by which they activated different segments of the trapezius muscle as the holding trial progressed.

De Luca (1985), however, has argued that there is no conclusive proof that the recruitment of additional motor units is the cause of increased amplitude in the EMG signal. He holds the view that the shift to lower MPF and the increase of RMS amplitude are in fact related, and they result from the increase in the low-frequency components of the myoelectrical signal, which means that more signal energy will be transmitted through the low-pass filtering effect of the body tissue. He also stated that the firing rates of the motor units do decrease throughout a sustained contraction, and since such decrease is more pronounced at the beginning of the exertion, the shift towards lower MPF will appear at this point, with the increase in the amplitude of the signal appearing later, towards the end of the exertion.

In fact, in some of the curves depicting the changes of relative RMS amplitude observed during the present study (shown in Appendix D), the change in the trapezius muscle appeared steeper towards the end of the holding, more noticeably on the left arm. However, even if the view held by

De Luca (1985) were the correct one, the question still remains as to why the two changes did not appear equally significant for both the deltoid and the trapezius muscles.

8.5.4 Relationship between EMG changes and subjective perception

As described in chapter 7 (section 7.4.5), the assessment of the linear relationship between the subjective (discomfort ratings) and objective (ΔEMG indices) indicators of fatigue led to the result of its ranging from the practically non-existent - the result for ΔMPF from the right trapezius muscle, with $R^2 = 0.2\%$- to just moderate, indicated by the value of 33.7% for ΔRMS amplitude from the left trapezius.

This result was rather unexpected, for a number of reasons. First, the relationship being searched was based on the underlying relationships between the change of discomfort ratings with the holding time, which had already been proven very strong (see chapter 6) and that between ΔEMG indices and holding times which, although when calculated for the whole of the data collected during trials at each abduction angle showed weak linearity (see table 7.8), in many individual trials it was fairly strong. Therefore, given the strength of the relationship between the subjective indicator of fatigue and the time, it was expected to influence substantially its relationship with the objective indicators.

Second, when only the data collected during the first trial by each subject at each abduction angle were analysed with view to their presentation during a congress (Serratos-Pérez and Haslegrave, 1994), the statistical treatment applied demonstrated the existence of linear relationships (albeit of moderate strength) for both ΔEMG indices in the six muscles studied.

Third, even though the issue of the relationship between subjective and objective indicators of fatigue has been the sole interest of a very limited number of studies, there is evidence that points to the existence of a significant correlation (Kilbom et al, 1983; Rohmert et al, 1986; Jørgensen et al, 1988; Hasson et al, 1989). However, the evidence is contradictory in regards of which are the variables correlated, and the nature of the relationship. Thus, Kilbom et al (1983) reported that the perceived effort - evaluated by the subject as a percentage of the effort expended during a previous trial of maximal endurance to handgrip applying 25% MVC- correlated well with blood pressure changes observed during sub-maximal trials at the same conditions. However, they found that the EMG changes did not correlate with the percentage rating of perceived effort. In contrast, Rohmert et al (1986) stated that ratings of perceived exertion (RPE, obtained using the 10-point scale of Borg, 1982) coincided with the EMG measurements obtained during trials in which young male subjects adopted several postures, two of them designed to tax the upper limbs, and sustained to exhaustion four different loads equivalent to proportions of between 25% and 75% of their MVC at that particular posture. The degree of coincidence was such that Rohmert et al affirmed that

RPE could replace the calculations of the load made employing EMG measurements. Jørgensen et al (1988) reported on a study in which the subjects performed intermittent pulling movements over a period of several hours during which the perceived exertion (also rated on Borg's 10-point scale) was related to the change in RMS amplitude. They found coefficients of correlation of 0.74 when the force applied was 20% MVC, and of 0.64 when it was 15% MVC, although the RPE ratings were only within the range between 2 and 5.

Even stronger evidence of a linear correlation between RPE (collected using Borg's 10-point scale) and EMG came from the study by Hasson et al (1989) who obtained their information from 10 male adults (average age 28.9 years, s.d. 2.1) who were asked to perform a handgrip at 50% MVC. The subjects had to apply that level of strength from the start, keeping it constant until they reached the endurance limit, defined as the moment when the subject could not sustain the target force for 3 consecutive seconds. The endurance limit ranged between 72.4 seconds and 103.2 seconds. EMG measurements and RPE ratings were obtained every 10 seconds. Applying linear regression technique, Hasson et al assessed the relationship between RPE and both MPF and RMS amplitude, using the average value of the variables. To get the most representative result, they considered only the measurements obtained up to the 80th second, so that all but the last averaged values included one per subject. The typical value of RPE at the start of the exertion was 3, and the calculations produced regression lines with values of R^2 between 0.802 and 0.958 (0.922 ±

0.016) for the relationship between RPE and MPF, and between 0.495 and 0.894 (0.729 ± 0.043) for that between RPE and RMS amplitude.

Thus, there was enough supporting evidence to expect the finding of a significant linear relationship in the present study. However, when the regression analysis on the data was performed, the results were a mixture (see table 7.10) of significant (highly so in some cases) slope coefficients, indicative of a significant linear relationship between the variables, with low (some cases extremely so) values of coefficient of determination, which meant that the fit between data and model was poor. The dominant trend present in the data is reflected by the value of R2, and this indicated that in most of the cases there was no underlying linear relationship between discomfort rating and EMG indices.

The discrepancy between a large level of significance for the coefficient slope and the extremely poor value of R^2 is due to the large number of data (1495 data pairs) fed into the process. This resulted in the numerator term of the quotient MSE_{model}/MSE_{error} being affected by only one degree of freedom, whilst the denominator term was rendered very small by having to divide the term SSE_{error} by a large number of degrees of freedom. Therefore, the conclusion must be reached that the information obtained from the process of fitting linear regression models to determine the presence and strength of the relationship under investigation was potentially misleading. Furthermore, the process of assessing that relationship has provided an insight into the kind of trouble that may be found when treating data since, due to the presence of conflicting statistical indicators, the results could have been interpreted in two exactly opposite ways.

8.5.5 Reversed changes in the EMG parameters and their implications

From the analysis of the myoelectric signals collected during the trials of postural exertion (presented in chapter 7), it was found that in a considerable number of cases the change in the EMG signal went in direction opposite to that usually associated with the presence of fatigue, that is MPF increased instead of decreasing and RMS amplitude decreased instead of increasing. Although such a phenomenon was present in all the six muscles studied, the reversed change of MPF was far more frequent in the trapezius muscles, and that of RMS amplitude appeared predominantly in the medial deltoids.

As explained in chapter 7, the significance of the changes in the EMG parameters, which was taken as indicator of fatigue, was assessed by testing whether their average value was different from zero. Judging by this criterion, a significant increase of MPF in the right trapezius was accompanied by a significant increase of RMS amplitude in 26 trials, and the same combination in the left trapezius occurred in 23 trials. Simultaneous significant decreases of RMS amplitude and MPF were seen in the right medial deltoid in 21 trials, and the left medial deltoid presented the same combination in 18 trials.

8.5.5.1 Load sharing and the reversed change of RMS amplitude

Rather surprisingly, the plots of the EMG changes over the holding time showed that, despite the large significance it exhibited in the majority of cases, the reduction of RMS amplitude in the medial deltoid muscles (marked with * in the graphs presented in Appendix D) did not appear to affect the nature of the changes in the posterior deltoid. This was indeed unexpected, since the evidence obtained during the two series of experiments performed in this study, both from the direct observation of the way the subjects tried to adjust the use of their muscles with the passage of the time, and from their reports about the sites of maximum discomfort, created the strong impression that the two portions of the deltoid were operating a mechanism of load sharing, so that any eventual reduction of the activity in the middle portion of the deltoid (evidenced by a reduction in RMS amplitude) would increase the load being borne by the posterior part, and this shift would show clearly in the signal from this muscle as a significant increase in its RMS amplitude.

A possible explanation of the absence of any conspicuous signs of load sharing between the portions of the deltoid muscles might be found in a report by Hagberg (1981a). He recorded the electrical activity from the descending part of trapezius, medial and anterior deltoids, biceps brachii, infraspinatus and supraspinatus (using intramuscular electrodes), during the abduction of the right arm at 90° to exhaustion by female subjects. He also observed a simultaneous decrease of both MPF and RMS amplitude in the medial deltoid, and proposed that the unexpected reduction of RMS amplitude was possibly due to a modification of the muscular function, by which the muscle transfers the torque acting on it to other muscles, most likely those of the rotator cuff. However, Hagberg did not back this assertion with any proof, despite having registered EMG signal from the supraspinatus muscle which is part of the rotator cuff and, given its location, might be the most likely source of relief for the deltoid.

Monod (1972, p 59) offered a very similar explanation for the reduction in the amplitude of the EMG signal, although without referring to any muscle in particular. He attributed it to a relief effect by the recruitment of fibres in the same muscle, rather than to the activation of other muscles. He proposed that during prolonged isometric contraction the subject can effect slight variations in the posture, which by bringing into action new non-fatigued muscular fasciculi, would take on the load from the portions of the muscle already heavily fatigued. Since these non-fatigued muscular units are in fact recruited from the deeper parts of the muscle, reducing the number of superficial active units, the electrodes placed on the surface would register it as a reduction in the amplitude of the signal, but intramuscular electrodes could still show an increase in the amplitude.

The decision taken from the onset of this investigation, that only non-invasive procedures were to be used in the collection of information, ruled out the use of intramuscular electrodes. Consequently, no EMG signal was recorded either from the muscles of the rotator cuff or from the deep layers of the superficial muscles studied. Therefore, the information collected in the course of the holding trials does not permit an assessment of the validity of the explanations to the simultaneous decrease of MPF and RMS amplitude offered by Hagberg (1981a) and by Monod (1972) .

8.5.5.2 Motor units recruitment in the reversed change of MPF

The simultaneous increase in MPF and in RMS amplitude for the trapezius muscle that occurred in the present study was also observed by Hagberg (1981a). This he attributed mainly to the recruitment by the trapezius muscle of additional, non-fatigued motor units, which at the level of strength involved in the abduction of the arm at 90° (less than 15-20% MVC) results not only in the increase of the amplitude of the signal, but also in the increase of MPF (Ericson and Hagberg, 1979).

Thus, the recruitment of additional, non-fatigued muscle fibres has been proposed as the explanation both for the reduction of RMS amplitude in circumstances where it was expected to increase (Monod, 1972) and for the increase of MPF when it should have decreased (Hagberg 1981a). Perhaps both phenomena may in fact be linked to a common root cause, and the final effect depends on the amount of force being developed by the muscle, as pointed out by Ericson and Hagberg (1979). Nevertheless, it is clear that much work still remains to be done in trying to achieve a complete understanding of the way muscles in the shoulder area respond to low-strength efforts sustained by a long time.

8.5.5.3 Possible causes for the reversed changes in the EMG parameters

The behaviour of the EMG parameters throughout the experiments performed in the present study, even assuming that the reversed changes of MPF and RMS amplitude may be explained through the mechanisms proposed by Hagberg (1981a), still leaves an important question without an obvious answer: what is it that determines that the EMG features will change in the unexpected direction during a certain trial? On the one hand, the combinations of increase of both parameters in the trapezius muscle, or their concurrent decrease in the deltoid muscles, were present in all the experimental conditions examined during the trials, they appeared at least once for every subject, and their occurrence was not traceable to variations in the experimental procedures. On the other hand, whilst the unexpected changes in the deltoid muscles appeared to be independent of the main experimental variables - that is they were not particularly linked to any abduction angle or to the gender of the subject, nor did they occur more often in one arm than in the other- the changes in the trapezius muscles did not seem to be completely free from the influence of the experimental factors, although such influence did not follow a defined pattern. So, they were more frequent in the right arm than in the left arm, at a 4:3 ratio; in the right arm they were nearly three times as frequent at 60° as at the other two angles, but practically of equal frequency among the female and the male subjects; in the left trapezius, however, the unexpected combination of changes was three times more frequent in the female subjects than in the males, but equally frequent at the three abduction angles.

Since all the trials were carried out in exactly the same way, and they were assigned to each subject in random order, the presence of the unexpected combinations of myoelectric changes may only be attributed to variations in the individual pattern of the responses, or perhaps they (the combinations) will have to be attributed entirely to chance, for the time being at least. Nevertheless, neither of the two suggestions may be seen as a fully satisfactory explanation and the issue demands further investigation since, although it has been previously reported by Hagberg (1981a) and by Takala and Viikari-Juntura (1991), the first study was in connection with the measurement of the endurance to the abduction at 90° (Hagberg, 1981a), and the second only used the short-term abduction at 90° with maximal strength as one of several tests in search for a relationship between the level of MVC among clerical workers and their liability to neck-shoulder musculoskeletal complaints. Apparently, the present investigation provides the first report about the existence of unusual electromyographic changes in a study comparing the effects of various postures.

8.5.6 Persistence of the signs of muscular fatigue.

EMG signals were collected during contractions performed in reference conditions 10 minutes before the start and 5 minutes after the completion of each of the trials conducted during the main experiment. A comparison of the features of those signals found that both MPF and RMS amplitude measured in the contraction post-exertion had returned to the level they were in during the measurement pre-exertion. This finding agreed with those of Petrofsky and Lind (1980), Mills (1982), Merletti et al (1983) and Kuorinka (1988), who have all reported that following the termination of a sustained contraction in which the frequency parameter of the EMG signals (either MPF or median frequency) showed a significant downward shift, it returned to well within its initial value in a time span of between 3 and 5 minutes.

However, all the studies mentioned above have only dealt with the behaviour of the frequency parameter. Kroon and Naeije (1991) studied the response of MPF, RMS amplitude, the rate of change of MPF to lower frequencies (d (MPF) / dt), the maximal strength of the muscle (MVC) and the endurance time during and after contractions of the left biceps muscle. The contractions were either isometric (at 50% MVC), concentric or eccentric (both at 40% MVC). MVC and endurance time were measured before the subjects completed the main experimental protocol, consisting in intermittent exertion, with 3s contraction and 2s rest, and 1 extra minute rest after each series of 10 contractions. The subjects had to sustain this work regime until reaching exhaustion. The value of d (MPF) / dt observed during the test of endurance prior to the completion of the main protocol, and the values of MPF and RMS amplitude at the beginning of it were recorded.

After the completion of the exertion to exhaustion, Kroon and Naeije (1991) monitored the five variables at set times over a period that extended for several days, until all five had returned to the values previously recorded. To do the monitoring, the subjects were asked to attempt the contractions at their MVC, and endurance (not exhaustion, note) trials were completed. MPF was the variable that returned the fastest to its pre-exertion value, for the three types of exertion it was already there when the first post-exertion measurement was obtained, 45 minutes after completion of the main experiment. For the rest of the variables, the longest lasting effects were provoked by the eccentric contraction, whilst the shortest were associated to the isometric mode. In this, the most relevant to the present investigation, MVC had returned to its pre-exertion level 90 minutes after the end of the main trial, d (MPF) / dt had done it within the third hour, but it took a full 2 days for RMS amplitude and the endurance time to get back to the reference, fatigue-free level.

There is a huge contrast between this result and the one obtained during the main experiment in the present investigation, where RMS amplitude had already gone back to its pre-trial value by the fifth minute after the completion of the holding. To understand the causes behind that contrast it is necessary to consider three fundamental differences between the present study and that of Kroon and Naeije's:

i) the mode of the exertion itself, with a continuous contraction by a large group of muscles in order to keep a static posture, against an intermittent contraction by a smaller muscle group, that of the elbow flexors;

ii) the amount of strength involved in the exertion. Whilst the value for the postural loading during abduction at 90° has been quoted as 20% of the middle deltoid's MVC (Hansson et al, 1992), 12.6% of the torque generated during arm abduction at MVC (Hagberg, 1981a) and between 13-18% of the same torque (Mathiassen and Winkel, 1991), the subjects in Kroon and Naeije's study applied 50% MVC.

iii) the extent and nature of the intramuscular modification that the two experimental protocols are likely to provoke, considering that whilst in the present study the subjects were required to sustain the exertion only to the point where they felt it was no longer bearable, Kroon and Naeije set the total exhaustion as the end point.

It is reasonable to assume that the disparity between the two studies, regarding the length of time it took for the EMG signs of fatigue to disappear, may be explained by the three factors mentioned above, particularly the third one.

The results of Kroon and Naeije (1991) also highlighted the fact that there is a gulf between the two possible criteria to judge the absence of fatigue following heavy exertion, for whilst those based on the return of the EMG parameters (MPF being perhaps the most frequently used) to their pre-fatigue values would have provided such indication just 45 minutes after the cessation of work, the measurement of the actual endurance to the task will only pronounce the 'all clear' after a full 48 hours. Therefore, if the decision as to when to reintegrate heavily fatigued workers to their tasks was taken based on the first criteria, that would actually mean their exposure to increasing levels of cumulative fatigue, which in turn could predispose them to long-term muscular injury. A similar reasoning had already been expressed by Funderburk et al (1974), who observed that the maximal strength with which a person is capable of performing a contraction recovers far quicker than their actual endurance to the exertion. Concretely, they found that whilst it took only 10 minutes to restore the subject's MVC following a series of static handgrip at levels of 20%, 40% or 60% MVC, not even after 40 minutes' rest had they fully retrieved their endurance capacity. In summary, the monitoring of recovery through the values of the EMG parameters or of the muscular strength on its own poses the risk of assuming that recovery has been achieved whilst in reality the capacity of the person to cope with the effort required from them is far from its optimum point.

8.5.7 Conclusions

The review of the results obtained from the analysis of the electromyographical information collected during the main experiment led to the following conclusions:

i) The holding of the postures with arms abducted resulted in myoelectric changes that indicate the presence of fatigue. However, unlike the changes of discomfort ratings, the EMG changes did not have a well-defined relationship with the holding time.

ii) The extent of the EMG changes observed in the present study was, in general, smaller than what other studies have reported, but no obvious reason for this fact could be found.

iii) There was a clear predominance of the changes of MPF in the deltoid muscles. In the trapezius muscles it was the change of RMS amplitude the one which predominated.

iv) No relationship was established between EMG changes and discomfort ratings.

v) There was a large number of cases with the EMG changes going in the direction opposite to that expected. Again, no obvious reason could be found for the existence of this unexpected feature. Neither was there a clear evidence of load sharing, when this was tested in relation with the appearance of reversed changes.

vi) Both MPF and RMS amplitude were back to their values pre-holding following five minutes' rest. Although this result agreed with most of the evidence already existent, there was a particularly interesting case of disagreement.

8.7 Suggestions for further work

Having completed the review of the experimental work carried out in the course of the present investigation, and the results it yielded, it is now possible to advance a number of suggestions regarding areas in which further work could prove both rewarding and successful in widening the knowledge basis in connection with isometric exertion, in its modality of purely postural work.

The first two of those areas to be mentioned represent in fact objectives originally set for the present investigation which could not be duly accomplished. First, there is the need to submit Milner's model to tests where male subjects are asked to complete a single combination work/rest/work to exhaustion, in at least one, but ideally the three postures with arms abducted that were investigated in the present study. Second, in order to fully replicate Milner's experimental approach, to incorporate into the tests the same form of 'secondary task' used during the development of the model, namely the playing of a video-game. However, it has to be borne in mind that such task must be implemented in a way such that it does not mean a significant departure from the conditions of pure postural loading that were created during the present investigation. Should the model prove viable under those conditions, that would then back the assertions made by this researcher regarding the strong influence of that factor on the results obtained by Milner, which probably flawed the whole process of construction of the model.

It is convenient to recall that these two original goals were set aside when the early experiments demonstrated the unsuitability of Milner's model to predict the recovery of the female subjects on which it was first tested. Getting back to them in due course was the finest intention of this researcher, but time limitations put paid to that.

The discussion of the results of the present study highlighted a number of possibilities for further work whose relevance rests mainly with their scientific (perhaps academic is also a suitable term) value. The following are those which this researcher, given the opportunity, would feel strongly inclined to pursue:

i) to extend the electromyographic study of postures with arm abduction beyond the single combination work/rest/work, with a view to finding out how would the accruing discomfort reflect on the myoelectric activity, particularly that of the trapezius muscle, whose role as the site for maximum discomfort was heightened by the performance of the second holding during the first series of trials carried out in the present study;

ii) to study the myoelectric activity generated by the holding of a wider array of work-related postures, probing for the existence of the phenomenon of reversed changes that was quite evident during the trials performed in this study, appearing in all the experimental conditions studied.

The experimental work also yielded a result of practical significance: the growth of the discomfort provoked by the pure postural exertion follows a strongly defined linear pattern, which exhibited the same shape regardless of the abduction angle and of the gender of the subject. The strength of this relationship is such that it may be turned into a model of the relationship between the passage of the holding time and the subjective reactions to this kind of exertion. As such, the model may then be used to predict either the degree of discomfort that might be expected to appear after the posture has been held for a certain time, or the maximum capacity for such task, knowing the rate of increase of discomfort associated with it.

However, in order to enhance the standing of such model from being valid only for the experimental conditions where it has emerged into becoming a generally applicable tool that may be confidently used in other set-ups it is first necessary to determine the maximum holding times for other working postures, particularly those where the upper body is subjected to considerable stress. In this regard, and to draw as much as possible from the findings of the present study, the first likely candidate would be the same postures with arm abduction included in it, only this time with the subject seated instead of standing up.

Another important issue in relation with the use of the model would be the search for the combination of anthropometric features and experimental factors (including the presence of external loads, which in the present study was foregone) that might lead to the construction of models with the ability to predict the maximum holding times for as many work-related postures as it was possible. This work would also serve to delve into the apparent discrepancy that in this regard exhibited the subjects who participated in the main experiment, for it was seen that, depending on the subject's gender, their anthropometric features tend to relate with the maximum length of the holding in different ways.

Naturally, the suggestions so far presented do not exhaust the possibilities that the researcher has been able to visualise, but they certainly are the most relevant. Also, no doubt about it, more possibilities will be revealed when the work is scrutinised further, hopefully not only by this researcher, but by many other Egonomists.

CHAPTER 9

CONCLUSIONS

The objectives set for the present investigation were stated in chapter 3. There were six specific objectives, which the experimental work - about to be described at that stage- strove to achieve. Those objectives will now be recalled one at a time, and the main conclusion drawn from the experimental work undertaken in pursuance of its completion will be expressed immediately afterwards.

Objective No. 1

To test the assertion made by Milner (1985) that a model to predict levels of remaining endurance to static postural work, developed from observations on a single standing and bent-forwards posture, is still valid when applied to other postures. In addition, since the model was derived from data obtained in a study of male subjects only, this research tested whether the model would also apply to female subjects.

Conclusion No. 1

The results obtained in the present study deny Milner's claims. The model he developed could not predict with acceptable accuracy the recovery to be achieved when the posture changed to an upright stance with both arms abducted at 60°. Besides, the evidence obtained suggests that the underlying assumptions on which Milner based the model are not as sound as he purported them to be. This observation significantly lessened the relevance of finding out whether the model behaves the same when tested on males and on females.

This task was set aside with the aim of completing it later but, due to the limited time frame in which this investigation had to be completed, such goal was not achieved.

Objective No. 2

To test how repeatable is the maximum holding time for a posture, which is assumed a valid indicator of the endurance to the loads created by that posture.

Conclusion No. 2

Being essentially an individual trait, the maximum capacity to hold a posture continuously exhibited the inherent variability that might be expected. This variability was in evidence not only when the feature was compared between subjects, but it also affected the repeated measurements on some of the subjects studied. Nevertheless, when the maximum holding time was evaluated as a group feature rather than at individual level, it was found to be a repeatable measure.

Objective No. 3

To evaluate the effects that postural variations and the gender of the subjects have on the maximum holding time.

Conclusion No 3

Increasing the abduction angle brought about a significant reduction in the maximum holding time, and it also decreased the dispersion of the values both at individual and at group level. On average, the capacity to endure postural loading of the shoulder was significantly larger among male subjects than it was among females, but a wide overlap between individuals was evident.

Objective No. 4

To establish how the subjective perceptions of fatigue develop during the course of maximum holding times, and the way they are affected by postural variations and gender of the subject.

Conclusion No. 4

The subjects perceived the discomfort to grow linearly. This linear relationship exhibited a remarkable strength which was not significantly affected by the change of the abduction angle, nor was it affected by the gender of the subject. Such strength and consistency open the possibility of using the linear relationship as a model to predict levels of maximum holding capacity based on the rate of increase of discomfort, or to predict the length of time it would take for the posture to provoke a given degree of discomfort. However, the interindividual variation of the holding time mean that the model would not function at individual level, it must be used to predict expected occurrences for groups of people.

Objective No. 5

To assess the presence of muscular fatigue as indicated by changes in the electromyographic signals, to investigate the influence of the experimental conditions on their nature and extent, and to look for the possible relationships between those changes and the subjective perception of fatigue.

Conclusion No. 5

The changes in the myoelectric activity that occurred during the maximal holding of the postures (determined via EMG analysis) clearly indicated the development of fatigue as consequence of the postural loading. There were differences in the extent of the changes which might be attributed to the influence of the abduction angle and of the gender of the subject; however, these did not follow a common and well defined pattern. Even though the results from many individual trials revealed the existence of a linear relationship between the myoelectric changes and the progress of the discomfort ratings provided by the subject, such linearity was no longer evident when the evaluation included the whole data. Another noticeable feature of the results was the presence of fairly large numbers of changes that went opposite to the direction usually associated with the presence of fatigue.

Objective No. 6

To assess the length of time over which the electromyographic signs of fatigue will persist following postural exertion of maximum duration.

Conclusion No. 6

The signs of fatigue evidenced by changes in the myoelectric activity had disappeared when EMG signals collected five minutes after the end of the exertion were analysed.

Thus, in bringing this chapter to a close - and with it the whole thesis- it is convenient to recall that there was an ulterior aim behind the pursuance of the six specific objectives just reviewed: to make a contribution to a research effort which ultimately aspires to eradicate, or at the very least reduce to its minimum expression, the presence of work-related harm to the musculoskeletal system. It may only be expected that the completion of the six subsidiary goals set to this investigation will in turn mean the fulfilment of that far-reaching aim.

REFERENCES

Aarås A. (1987) Postural load and the development of musculo-skeletal illness. Scandinavian Journal of Rehabilitation Medicine, Supplement No. 18: 1-35

Aarås A., Westgaard R.H. (1987) Further studies of postural load and musculo-skeletal injuries of workers at an electromechanical assembly plant. Applied Ergonomics, 18: 211-219.

Andersson G.B.J., Örtengren R. (1984) Assessment of back load in assembly line work using electromyography. Ergonomics, 27: 1157-1168.

Arendt-Nielsen L., Mills R.K., Forster A. (1989) Changes in muscle fibre conduction velocity, mean power frequency, and mean EMG voltage during prolonged submaximal contractions. Muscle and Nerve, 12: 493-497.

Barbonis P.A. (1979) Measuring and Modelling of Postural Work Load. PhD thesis, University of Birmingham.

Basmajian J.V., De Luca C.J. (1985) Muscles Alive: their functions revealed by electromyography. Fifth edition. Baltimore: William and Wilkins.

Bigland-Ritchie B., Donovan E.F., Roussos C.S. (1981) Conduction velocity and EMG power spectrum changes in fatigue of sustained maximal efforts. Journal of Applied Physiology, 51: 1300-1305.

Bjelle A., Hagberg M., Michaelsson G. (1979) Clinical and ergonomic factors in prolonged shoulder pain among industrial workers. Scandinavian Journal of Work, Environment and Health, 5: 205-210.

Bjelle A., Hagberg M., Michaelsson G. (1981) Occupational and individual factors in acute shoulder-neck disorders among industrial workers. British Journal of Industrial Medicine, 38: 356-363.

Bjøkrsten M., Jonsson B. (1977) Endurance limit of force in long-term intermittent static contractions. Scandinavian Journal of Work, Environment and Health, 3: 23-27.

Blache K.M., Dickens K., Pappas P. (1987) Postural discomfort evaluation of production workers. In: Asfour S.S. (ed) Trends in Ergonomics/Human Factors IV: 655-662. Amsterdam: North-Holland.

Blinowska A.J., Verroust J., Cannet G. (1980) An analysis of synchronization and double discharge effects on low frequency electromyographic power spectra. Electromyography and Clinical Neurophysiology, 20: 465-480.

BMDP (1988) BMDP Statistical Software Manual, v 1. BMDP Statistical Software Inc. Los Angeles: UCLA Press.

Bonney R., Weisman G., Haugh L.D., Finkelstein J. (1990) Assessment of postural discomfort. Proceedings of the HFS 34th Annual Meeting, pp 684-687.

Borg G.A.V. (1973) A note on a category scale with 'ratio properties' for estimating perceived exertion. Report No. 36. The Institute of Applied Psychology, University of Stocholm.

Borg G. (1982) A category scale with ratio properties for intermodal and interindividual comparisons. In: Geissler H.G., Petzold P. (eds) Psychophysical Judgment and the Process of Perception, pp 25-34. Amsterdam: North-Holland.

Borg G. (1990) Psychophysical scaling with applications in physical work and the perception of exertion. Scandinavian Journal of Work, Environment and Health, 16 (supplement 1): 55-58.

Boussenna M., Corlett E.N., Pheasant S. (1982) The relation between discomfort and postural loading at the joints. Ergonomics, 25: 315-322.

Byström S.E.G., Kilbom Å. (1990) Physiological response in the forearm during and after isometric intermittent handgrip. European Journal of Applied Physiology, 60: 457-466.

Caffier G., Heinecke D., Hinterthan R. (1993) Surface EMG and load level during long-lasting static contractions of low intensity. International Journal of Industrial Ergonomics, 12: 77-83.

Caldwell L.S., Smith R.P. (1966) Pain and endurance of isometric muscle contractions. Journal of Engineering Psychology, 5: 25-32.

Chaffin D.B. (1973) Localized muscle fatigue - definition and measurement. Journal of Occupational Medicine, 15: 346-354.

Christensen H. (1986) Muscle activity and fatigue in the shoulder muscles of assembly-plant employees. Scandinavian Journal of Work, Environment and Health, 12: 582-587.

Corlett E.N. (1981) Pain, posture and performance. In: Corlett E.N., Richardson J. (eds) Stress, Work Design and Productivity, pp 27-42. London: Wiley.

Corlett E.N. (1988) The investigation and evaluation of work and workplaces. Ergonomics, 31: 727-734.

Corlett E.N., Bishop R.P. (1976) A technique for assessing postural discomfort. Ergonomics, 19: 175-182.

Corlett E.N., Manenica I. (1980) The effects and measurement of working postures. Applied Ergonomics, 11: 7-16.

Corlett E.N., Manenica I., Goillau P.J. (1983) The relationship between EMG activity of the sacrospinalis and reported back discomfort. European Journal of Applied Physiology, 50: 213-222.

Daanen H.A.M., Mazure M., Holewijn M., Van der Velde E.A. (1990) Reproducibility of the mean power frequency of the surface electromyogram. European Journal of Applied Physiology, 61: 274-277.

Davies B.T., Pratt M.C. (1976) The effect of static work on a subsequent fine manipulation task. Proceedings of the 6th IEA Congress, pp 531-534.

Deeb J.M., Drury C.G. (1990) A methodology for muscular isometric endurance and fatigue research. International Journal of Industrial Ergonomics, 6: 255-260.

Deeb J.M., Drury C.G., Pendergast D.R. (1992) An exponential model of isometric muscular fatigue as a function of age and muscle groups. Ergonomics, 35: 899-918.

Dieën (van) J.H., Toussaint H.M., Thissen C., van de Ven A. (1993) Spectral analysis of erector spinae EMG during intermittent isometric fatiguing exercise. Ergonomics, 36: 407-414.

Donald K.W., Lind A.R., McNicol G.W., Humphreys P.W., Taylor S.H., Staunton H.P. (1967) Cardiovascular responses to sustained (static) contractions. Circulation Research, 20-21 (supplement 1): 15-32.

Douwes M., Dul J. (1993) Studies on the validity of a work-rest model for static postures. In: Marras W.S. et al (eds) The Ergonomics of Manual Work, pp 341-343

Drury C.G., Spitz G. (1978) Strength, duration and recovery mechanisms. In: Safety in Manual Materials Handling. Report on International Symposium. State University of New York at Buffalo, July 18-20, 1976. DHEW (NIOSH) Publication No. 78-185.

Dul J., Douwes M., Smitt P. (1990) A work-rest model for static postures. In: Högfors C. (ed) Proceedings from the 1990 Biomechanics Seminar. Goteborg: Chalmers University of Technology.

Dul J., Douwes M., Smitt P. (1991) A work-rest model for static postures. In: Quéinnec Y., Daniellou F (eds) Designing for everyone. Proceedings 11th Congress IEA, v1: 93-95. London: Taylor & Francis.

Dul J., Douwes M., Miedema M. (1993) A guideline for the prevention of discomfort of static postures. In: Nielsen R, Jørgensen K. (eds) Advances in Industrial Ergonomics and Safety V: pp 3-5. London: Taylor & Francis.

Dul J., Douwes M., Smitt P. (1994) Ergonomic guidelines for the prevention of discomfort of static postures based on endurance data. Ergonomics, 37: 807-815.

Eason R.G. (1960) Electromyographic study of local and generalized muscular impairment. Journal of Applied Physiology, 15: 479-482.

Edwards R.G., Lippold O.C. (1956) The relation between force and integrated electrical activity in fatigued muscle. Journal of Physiology (London) 132: 677-681.

Edwards R.H.T. (1981) Human muscle function and fatigue. In: Porter R., Whelan J. (eds) Human Muscle Fatigue: Physiological Mechanisms. Ciba Foundation Symposium 82. London: Pitman Medical.

Ericson B.E., Hagberg M. (1979) EMG power spectra versus muscular contraction level. In: Persson A. (ed) Abstracts from the 6th International Congress of Electromyography. Acta Neurologica Scandinavica, Supplement 73, 163.

Fallentin N., Jørgensen K. (1992) Blood pressure response to low level static contractions. European Journal of Applied Physiology, 64: 455-459.

Fine L., Silverstein B., Armstrong T., Joseph B., Buchholz B. (1986) A pilot study of postural characteristics of jobs associated with an elevated risk of rotor cuff tendinitis. In: Corlett E.N., Wilson J., Manenica I. (eds) The Ergonomics of Working Postures, pp 39-43. London: Taylor and Francis.

Gaffney F.A., Sjøgaard G., Saltin B. (1990) Cardiovascular and metabolic responses to static contraction in man. Acta Physiologica Scandinavica, 138: 249-258.

Genaidy A.M., Karwowsky W. (1993) The effects of neutral posture deviations on perceived joint discomfort ratings in sitting and standing postures. Ergonomics, 36: 785-792.

Gerdle G., Eriksson N.E., Hagberg C. (1988) Changes in the surface electromyogram during increasing isometric shoulder forward flexion. European Journal of Applied Physiology, 57: 404-408.

Grandjean E., Hünting W. (1977) Ergonomics of posture -review of various problems of standing and sitting posture. Applied Ergonomics, 8: 135-140.

Grinten (van der) M.P. (1991) Test-retest reliability of a practical method for measuring body part discomfort. In: Quéinnec Y., Daniellou F (eds) Designing for everyone. Proceedings 11th Congress IEA, v1: 54-56. London: Taylor & Francis.

Hagberg M. (1981a) Electromyographic signs of shoulder muscular fatigue in two elevated arm positions. American Journal of Physical Medicine, 60: 111-121.

Hagberg M. (1981b) Muscular endurance and surface electromyogram in isometric and dynamic exercise. Journal of Applied Physiology, 51: 1-7.

Hagberg M. (1982) Local shoulder muscular strain - symptoms and disorders. Journal of Human Ergology, 11: 99-108.

Hagberg M. (1984) Occupational stress and disorders of the neck and shoulder: a review of possible pathophysiology. International Archives of Occupational and Environmental Health, 53: 269-278.

Hagberg M., Michaelson G., Õrtelius A. (1982) Serum creatine kinase as an indicator of local muscular strain in experimental and occupational work. International Archives of Occupational and Environmental Health, 50: 377-386.

Hägg G.M. (1991) Comparison of different estimators of electromyographic spectral shifts during work when applied on short test contractions. Med Biol Eng Comput 29: 511-516.

Hägg G.M. (1992) Interpretation of EMG spectral alterations and alteration indexes at sustained contraction. Journal of Applied Physiology, 73: 1211-1217.

Hansson G.Å, Strömberg U., Larsson B., Ohlsson K., Balogh I, Moritz U. (1992) Electromyographic fatigue in neck/shoulder muscles and endurance in women with repetitive work. Ergonomics, 35: 1341-1352.

Harms-Ringdahl K., Brodin H., Eklund L., Borg G. (1983) Discomfort and pain from loaded passive joint structures. Scandinavian Journal of Rehabilitation Medicine, 15: 205-211.

Hasson S.M., Signorile J.F., Williams J.H. (1989) Fatigue-induced changes in myoelectric signal characteristics and perceived exertion. Canadian Journal of Sport Science, 14: 99-102.

Herberts P., Kadefors R., Broman H. (1980) Arm positioning in manual tasks. An electromyographic study of localized muscle fatigue. Ergonomics, 23: 655-665.

Hermansen L., Osnes J.B. (1972) Blood and muscle pH after maximal exercise in man. Journal of Applied Physiology, 32: 304-309.

Humphreys P.W., Lind A.R. (1963) Blood flow through active and inactive muscles of the forearm during sustained handgrip contractions. Journal of Physiology, 166: 120-135.

Janda V. (1983) Muscle Function Testing. London: Butterworths.

Jensen B.R., Schibye B., Sjøgaard K., Simonsen E.B., Sjøgaard G. (1993) Shoulder muscle load and muscle fatigue among industrial sewing-machine operators. European Journal of Applied Physiology, 67: 467-475.

Jones D.A., Round J.M. (1990) Skeletal Muscle in Health and Disease. Manchester: Manchester University Press.

Jones N.B., Lago P.J.A. (1982) Spectral analysis and the interference EMG. IEEE Proceedings, 129: 673-678.

Jonsson B. (1978) Kinesiology - with special reference to electromyographic kinesiology. In: Cobb W.A., van Dujin H. (eds) Contemporary Clinical Neurophysiology, Supplement 34: 417-428. Amsterdam: Elsevier.

Jonsson B. (1982) Measurement and evaluation of local muscular strain in the shoulder during constrained work. J. Human Ergology, 11: 73-88.

Jonsson B., Hagberg M. (1974) The effect of different working heights on the deltoid muscle. A preliminary methodological study. Scandinavian Journal of Rehabilitation Medicine, 3: 26-32.

Jørgensen K., Fallentin N., Krogh-Lund C., Jensen B. (1988) Electromyography and fatigue during prolonged, low-level static contraction. European Journal of Applied Physiology, 57: 316-321.

Jørgensen K., Fallentin N., Sidenius B. (1989) The strain on the shoulder and neck muscles during letter sorting. International Journal of Industrial Ergonomics, 3: 243-248.

Kadefors R., Kaiser E., Petersén I. (1968) Dynamic spectrum analysis of myo-potentials with special reference to muscle fatigue. Electromyography, 8: 39-51.

Kadefors R., Petersén I., Herberts P. (1976) Muscular reaction to welding work: an electromyographic investigation. Ergonomics, 19: 543-558.

Kahn J.F., Monod H. (1989) Fatigue induced by static work. Ergonomics, 32: 839-846.

Keyserling W.M., Punnett L., Fine L.J. (1987) Postural stress of the trunk and shoulders: identification and control of occupational risk factors. In: Ergonomic Interventions to prevent musculoskeletal Injuries in Industry. (ACGIH) Chelsea, Mich., USA: Lewis Publishers.

Kilbom Å., Persson J. (1981) Circulatory response to static muscle contractions in three different muscle groups. Clinical Physiology, 1: 215-225.

Kilbom Å., Gamberale F., Persson J., Annwall G. (1983) Physiological and psychological indices of fatigue during static contractions. European Journal of Applied Physiology, 50: 179-193.

Kilbom Å., Persson J., Jonsson B. (1986) Risk factors for work related disorders of the neck and shoulder -with special emphasis on working postures and movements. In: Corlett E.N., Wilson J.R., Manenica I. (eds), The Ergonomics of Working Postures, pp 44-53. London: Taylor & Francis.

Kinsman R.A., Weiser P.C. (1976) Subjective symptomatology during work and fatigue. In: Simonson E., Weiser P.C. (eds) Psychological Aspects and Physiological Correlates of Work and Fatigue, pp 336-403.

Kippers V., Parker A.W. (1984) Posture related to myoelectrical silence of erectores spinae during trunk flexion. Spine, 9: 740-745.

Kirk N.S., Sadoyama I. (1973) A relationship between endurance and discomfort in static work. MSc Report. Department of Human Sciences, Loughborough University of Technology.

Kogi K, Hakamada T. (1962) Slowing of surface electromyogram and muscle strength in muscle fatigue. Reports of the Institute of Science of Labour, 60: 27-41.

Komi P.V., Tesch P. (1979) EMG frequency spectrum, muscle structure and fatigue during dynamic contractions in man. European Journal of Applied Physiology, 42: 41-50.

Krogh A., Lindhard J. (1913) The regulation of respiration and circulation under the initial stages of muscular work. Journal of Physiology (London) 47: 112-136.

Krogh-Lund C. (1993) Myo-electric fatigue and force failure from submaximal static elbow flexion sustained to exhaustion. European Journal of Applied Physiology, 67: 389-401.

Krogh-Lund C., Jørgensen K. (1991) Changes in conduction velocity, medial frequency, and root mean square-amplitude of the electromyogram during 25% maximal voluntary contraction of the tricpes brachii muscle, to limit of endurance. European Journal of Applied Physiology, 63: 60-69.

Krogh-Lund C., Jørgensen K. (1992) Modification of myoelectric power spectrum in fatigue from 15% maximal voluntary contraction of human elbow flexor muscles to limit of endurance: reflection of conduction velocity variation and/or centrally mediated mechanisms? European Journal of Applied Physiology, 64: 359-370.

Krogh-Lund C., Jørgensen K. (1993) Myo-electric fatigue manifestations revisited: power spectrum, conduction velocity, and amplitude of human elbow flexor muscles during isolated and repetitive endurance contractions at 30% maximal voluntary contraction. European Journal of Applied Physiology, 66: 161-173.

Kroll W. (1968) Isometric fatigue curves under varied intertrial recuperation periods. Research Quarterly, 39: 106-115.

Kronberg M., Németh G., Broström L-A. (1990) Muscle activity and coordination in the normal shoulder. An electromyographic study. Clinical Orhopedia and Related Research, 257: 76-85.

Kroon G.W., Naeije M. (1991) Recovery of the human biceps electromyogram after heavy eccentric, concentric or isometric exercise. European Journal of Applied Physiology, 63: 444-448.

Kuorinka I. (1981) Discomfort and motor skills in semi-paced tasks. In: Salvendy G., Smith M. (eds) Machine Pacing and Occupational Stress, pp 295-301. London: Taylor & Francis. ✓

Kuorinka I. (1983) Subjective discomfort in a simulated repetitive task. Ergonomics, 26: 1089-1101.

Kuorinka I. (1988) Restitution of EMG spectrum after muscular fatigue. European Journal of Applied Physiology, 57: 311-315.

Lee K.S. (1987) Application of electromyography for musculoskeletal stress evaluation in sedentary work. In: Asfour S.S. (ed) Trends in Ergonomics/Human Factors IV, pp 693-700. Amsterdam: Elsevier Science Publishers.

Lind A.R., Mc Nicol G.W. (1967) Circulatory responses to sustained handgrip contractions performed during other exercise, both rhythmic and static. Journal of Physiology, 192: 595-607.

Lindström L., Magnusson R., Petersén I. (1970) Muscular fatigue and action potential conduction velocity changes studied with frequency analysis of EMG signals. Electromyography, 10: 341-355.

Lindström L., Kadefors R., Petersén I. (1977) An electromyographic index for localized muscle fatigue. Journal of Applied Physiology, 43: 750-754.

Lindström L., Petersén I. (1983) Electromyography in muscle fatigue studies: power spectrum analysis and signal theory aspects. In: Knuttgen, H.G., Vogel J.A., Poortmans J. (eds) Biochemistry of Exercise. International Series on Sport Sciences, 13: 187-196. Champaigne, Il: Human Kinetics.

Lippold O.C.J., Redfearn J.W.T., Vuco J. (1960) The electromyography of fatigue. Ergonomics, 3: 121-131.

Lloyd A.J. (1971) Surface electromyography during sustained isometric contractions. Journal of Applied Physiology, 30: 713-720.

Luca (De) C.J. (1985) Myoelectrical manifestations of localized muscular fatigue in humans. CRC Critical Reviews in Biomedical Engineering, 11: 251-279.

Luca (De) C.J., Knaflitz M. (1992) Surface Electromyography: What's New? Torino: Edizioni C.L.U.T.

Lucas D.B. (1973) Biomechanics of the shoulder joint. Archives of Surgery, 107: 425-432.

Maeda K. (1977) Occupational cervicobrachial disorders and its causative factors. Journal of Human Ergology, 6: 193-202.

Manenica I. (1986) A technique for postural load assessment. In: Corlett E.N., Wilson J., Manenica I. (eds) The Ergonomics of Working Postures, pp 270-277. London: Taylor & Francis.

Marras W.S. (1987) Preparation, recording and analysis of the EMG signal. In: Asfour S.S. (ed) Trends in Ergonomics/Human Factors IV, pp 701-707. Amsterdam: Elsevier Science Publishers.

Marras W.S. (1990) Industrial electromyography (EMG). Guidelines. International Journal of Industrial Ergonomics, 6: 89-93.

Mathiassen S.E., Winkel J. (1991) Cardiovascular responses during and after intermittent isometric shoulder-neck exercises. In: Quéinnec Y., Daniellou F (eds) Designing for everyone. Proceedings 11th Congress IEA v1: 126-128. London: Taylor & Francis.

Mathiassen S.E., Winkel J. (1992) Can occupational guidelines for work-rest schedules be based on endurance time data? Ergonomics, 35: 253-259.

Maton B. (1981) Human motor unit activity during the onset of muscle fatigue in submaximal isometric isotonic contraction. European Journal of Applied Physiology, 46: 271-281.

McArdle W.D. Katch F.I., Katch V.L. (1991) Exercise Physiology. Third edition. London: Lea & Febiger.

Merletti R., Sabbahi M.A., De Luca C.J. (1983) Median frequency of the myoelectric signal: effects of ischemia and cooling. European Journal of Applied Physiology, 52: 258-268.

Menzer J.A., Smith R.P., Caldwell L.S. (1969) Self-paced and irregular methods of subjective estimation of pain. Psychonomic Science, 15: 287-288.

Mills K.R. (1982) Power spectral analysis of electromyogram and compound muscle action potential during muscle fatigue and recovery. Journal of Physiology, 326: 401-407.

Miller I., Freund J.E. (1977) Probability and Statistics for Engineers. London: Prentice-Hall International Inc.

Milner N.P. (1985) Modelling Fatigue and Recovery in Static Postural Exercise. PhD thesis, University of Nottingham.

Milner N.P., Corlett E.N., O' Brien C. (1986) A model to predict recovery from maximal and submaximal isometric exercise. In: Corlett E.N., Wilson J., Manenica I. (eds) The Ergonomics of Working Postures, pp 126-135. London: Taylor & Francis. ✓

MINITAB (1991) MINITAB Reference Manual, pp 7-1 to 7-27. Minitab Inc., State College, PA.

Mirka G.A. (1991) The quantification of EMG normalization error. Ergonomics, 34: 343-352.

Misner J.E., Going S.B., Massey B.H., Ball T.E., Bemben M.G., Essandoh L.K. (1990) Cardiovascular response to sustained maximal voluntary static muscle contraction. Medicine and Science in Sports and Exercise, 22: 194-199.

Monod H. (1972) How muscles are used in the body. In: Bourne G.H. (ed) The Structure and Function of Muscle, 2nd Ed., Vol 1, Part 1: 23-74. New York: Academic Press.

Monod H., Scherrer J. (1964) Consequences ergonomiques de certaines données fondamentales concernant la contraction statique. Proceedings of the 2nd IEA Congress, Dortmund, Ergonomics, 45-52.

Monod H., Scherrer J. (1965) The work capacity of a synergic muscular group. Ergonomics, 8: 329-338.

Moritani T., Nagata A., Muro M. (1982) Electromyographic manifestations of muscular fatigue. Medicine and Science in Sports and Exercise, 14: 198-202.

Neter J., Wasserman W. (1977) Applied Linear Statistical Models, first edition, seventh printing. Homewood, Ill (USA) Richard D. Irwin, Inc.

Nutter D.O., Schlant R.C., Hurst J.W. (1972) Isometric exercise and the cardiovascular system. Modern Concepts of Cardiovascular Disease, 41: 11-15.

Öberg T., Sandjö L., Kadefors R. (1990) Electromyogram mean power frequency in non-fatigued trapezius muscle. European Journal of Applied Physiology, 61: 362-369.

Örtengren R, Andersson G., Bromann H., Magnusson R., Petersén I. (1975) Vocational electromyography: studies of localized muscle fatigue in the assembly line. Ergonomics, 18: 157-174.

Owen C.A. (1969) The painful shoulder. Journal of Occupational Medicine, 11: 85-90.

Paterson W. D. (1928) Circulatory and respiratory changes in response to muscular exertion in man. Journal of Physiology (London), 66: 323-345.

Perry, J. (1988) Muscle control of the shoulder. In Rowe C.R. (ed) The Shoulder, chapter 2: 17-34. New York: Churchill-Livingstone.

Petrofsky J.S., Lind A.R. (1980) The influence of temperature on the amplitude and frequency components during brief and sustained isometric contractions. European Journal of Applied Physiology, 44: 189-200.

Pheasant, S. (1986) Bodyspace. Anthropometry, Ergonomics and Design. London: Taylor & Francis.

Putz-Anderson V. (1988) (ed) Cumulative Trauma Disorders - A Manual for Musculoskeletal Diseases of the Upper Limbs. London: Taylor and Francis.

Roaf R. (1977) Posture. London: Academic Press

Rohmert W. (1960) Ermittlung von erholungspausen für statsiche arbeit des menschen. Internationale Zeitschrift für angewandte Physiologie einschliesslig Arbeitsphysiologie, 18: 123-164.

Rohmert W. (1965) Physiologische grunglagen der erholungszeitbestimmung. Zbl Arbeit Wiss, 19: 1-12.

Rohmert W. (1973) Problems in determining rest allowances. Part 1: use of modern methods to evaluate stress and strain in static muscular work. Applied Ergonomics, 4: 91-95.

Rohmert W., Wangenheim M., Mainzer J., Zipp P., Lesser W. (1986) A study stressing the need for a static postural force model for work analysis. Ergonomics, 29: 1235-1249.

Sadoyama T., Masuda T., Miyano H. (1983) Relationship between muscle fibre conduction velocity and frequency parameters of surface EMG during sustained contraction. European Journal of Applied Physiology, 51: 247-256.

Sahlin K., Harris R.C., Hultman E. (1975) Creatine kinase equilibrium and lactate content compared with muscle pH in muscle samples obtained after isometric exercise. Biochemical Journal, 152: 173-181.

Sällström J., Schmidt H. (1984) Cervicobrachial disorders in certain occupations, with special reference to the compression in the thoracic outlet. American Journal of Industrial Medicine, 6: 45-52.

Sakakibara Y., Honda Y. (1990) Cardiopulmonary responses to static exercise. Annals of Physiological Anthropology, 9: 153-161.

Sato H., Ohashi J., Iwanaga K., Yoshitake R., Shimada K. (1984) Endurance time and fatigue in static contractions. Journal of Human Ergology, 13: 147-154. ✓

Schierhout G.H., Myers J.E., Bridger R.S. (1993) Musculoskeletal pain and ergonomic stressors in manufacturing industry in South Africa. International Journal of Industrial Ergonomics, 12: 3-11.

Schüldt K., Ekholm J., Harms-Ringdahl K., Nemeth G., Arborelius U.P. (1986) Effects of changes in sitting working posture on static neck and shoulder muscle activity. Ergonomics, 29: 1525-1537.

Serratos-Pérez J.N. (1990) Health and Safety and Ergonomics in Workplaces of the Clothing Industry. Msc Thesis. Loughborough University of Technology.

Serratos-Pérez J.N., Haslegrave C.M. (1992) Modelling fatigue and recovery in working postures. In: Lovesey E.J. (ed) Contemporary Ergonomics 1992, 66-71. London: Taylor and Francis.

Serratos-Pérez J.N., Mendiola-Anda C. (1993) Musculoskeletal disorders among male sewing machine operators in shoemaking. Ergonomics, 36: 793-800.

Serratos-Pérez J.N., Haslegrave C.M. (1994) Relationship between subjective perception and EMG signs of muscular fatigue in shoulder-loading postures. Accepted for presentation at the 12th Triennial Congress of the International Ergonomics Association. Toronto, August 1994.

Sigholm G., Herberts P., Alström C., Kadefors R. (1984) Electromyographic analysis of shoulder muscle load. Journal of Orthopaedics Research, 1: 379-386.

Sjøgaard G. (1986) Intramuscular changes during long-term contraction. In: Corlett E.N., Wilson J., Manenica I. (eds) The Ergonomics of Working Postures, pp 136-143. London: Taylor and Francis.

Sjøgaard G., Savard G., Juel C. (1988) Muscle blood flow during isometric activity and its relation to muscle fatigue. European Journal of Applied Physiology, 57: 327-335.

Sommerich C.M., McGlothlin J.D., Marras W.S. (1993) Occupational risk factors associated with soft tissue disorders of the shoulder: a review of recent investigations in the literature. Ergonomics, 36: 697-717.

Stranden E., Åaras A., Anderson D.M., Myhre H.O., Martinsen K. (1983) The effects of working postures on muscular-skeletal load and circulatory condition. In: Coombes K. (ed) Proceedings of the 1983 Ergonomics Society Conference, pp 163-167. London: Taylor & Francis.

Strasser H., Keller E., Müller K.W., Ernst J. (1989) Local muscular strain dependent on the direction of horizontal arm movements. Ergonomics, 32: 899-910.

Svensson O.K., Arborelius U.P., Ekholm J. (1987) Relative mechanical load on shoulder and elbow muscles in standing position when handling materials manually. A study of packing work. Scandinavian Journal of Rehabilitation Medicine 19: 169-178.

Takala E-P, Viikari-Juntura E. (1991) Muscle force, endurance and neck-shoulder symptoms of sedentary workers. An experimental study on bank cashiers with and without symptoms. International Journal of Occupational Ergonomics, 7: 123-132.

Takala E-P., Lammi S., Nieminen H., Viikari-Juntura E. (1993) Electromyographic changes in the static holding test of the arm. International Journal of Industrial Ergonomics, 12: 85-90. ✓

Tesch P., Sjödin B., Thorstensson A., Karlsson J. (1978) Muscle fatigue and its relation to lactate accumulation and LDH activity in man. Acta Physiologica Scandinavica, 103: 413-420.

Tracy M.F. (1990) Biomechanical methods in posture analysis. In: Wilson J.R. and Corlett E.N. (eds) Evaluation of Human Work. pp 571-604. London: Taylor & Francis.

Tuttle W.W., Horvath S.M. (1957) Comparison of effects of static and dynamic work on blood pressure and heart rate. Journal of Applied Physiology, 10: 294-296.

Veiersted K.B. (1991) The reproducibility of test contractions for calibration of electromyographic measurements. European Journal of Applied Physiology, 62: 91-98.

Viitasalo J.T., Komi P.V. (1977) Signal characteristics of EMG during fatigue. European Journal of Applied Physiology, 37: 111-121.

Viitasalo J.T., Komi P.V. (1978) Isometric endurance, EMG power spectrum and fiber composition in human quadriceps muscle. In: Asmussen E., Jørgensen K. (eds) Biomechanics VI-A: 244-250. Baltimore: University Park Press.

Viitasalo J.T., Komi P.V. (1980) EMG, reflex and reaction time components, muscle structure and fatigue during intermittent isometric contractions in man. International Journal of Sports Medicine, 1: 185-190.

Wallace M., Buckle P. (1987) Ergonomic aspects of neck and upper limb disorders. In: Oborne D.J. (ed) International Reviews of Ergonomics, vol. 1: 173-200. London: Taylor & Francis.

Wely (van) P. (1970) Design and disease. Applied Ergonomics, 1: 262-269.

Westgaard R.H., Aarås A. (1984) Postural muscle strain as a causal factor in the development of musculoskeletal illness. Applied Ergonomics, 15: 162-174.

Westgaard R.H., Waersted M., Janssen T., Aarås A. (1986) Muscle load and illness associated with constrained body postures. In: Corlett E.N., Wilson J.R. and Manenica I. (eds) The Ergonomics of Working Postures. London: Taylor & Francis.

Westgaard R.H. (1988) Measurement and evaluation of postural load in occupational work situations. European Journal of Applied Physiology, 57: 291-304.

Westgaard R.H., Veiersted K.B., Andersen P. (1991) The development of occupational musculoskeletal disorders in the shoulder and neck: a longitudinal study. In: Quéinnec Y., Daniellou F. (eds) Designing for Everyone, Proc 11th Congress IEA, v1: 132-134.

Wiker S.F., Chaffin D.B., Longolf G.D. (1989) Shoulder posture and localized muscle fatigue and discomfort. Ergonomics, 32: 211-237.

Wiker S.F., Chaffin D.B., Longolf G.D. (1990) Shoulder postural fatigue and discomfort. International Journal of Industrial Ergonomics, 5: 133-146.

Winter D.A., Rau G., Kadefors R., Broman H., De Luca C.J. (1980) Units, terms and standards in the reporting of EMG research. Report by the ad-hoc Committee of the International Association of Electromyographic Kinesiology.

Appendix A

Information package handed to the subjects and other printed materials used in the experiments

NOTE: A copy of the body map and Borg's scale were also handed to the subjects. Since these were shown in chapter 3 when describing the methodology, they are not included here.

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

Information Package

INSTITUTE FOR OCCUPATIONAL ERGONOMICS

DEPARTMENT OF MANUFACTURING ENGINEERING
AND OPERATIONS MANAGEMENT

UNIVERSITY OF NOTTINGHAM

September 1991

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

Basic Information

This research project is located in the realm of occupational ergonomics; its ultimate goal is a contribution to improve the conditions in which an important number of workers perform their jobs. The aim of the experiments we are about to undertake is to obtain a better understanding of the way fatigue develops when a person adopts a given posture, which has been observed to occur in actual work situations.

The experimental posture involves standing with both upper arms partially raised to the sides, in line with your body; elbows flexed in a right angle; forearms parallel to the floor; each hand touching a small piece of a plastic sheet located in front of you. The whole experiment consists of 10 sessions to be performed over a period of four weeks and in all but the first of them you will be asked to adopt the described posture and hold it for as long as you can; until you have to stop through fatigue. This may involve a certain degree of discomfort, but you may at any time withdraw from taking part in the experiments if you wish.

Each experimental session will last around 30 minutes and the times may be arranged to suit your needs. You will be paid at a rate of £1.50 per session and the money will be handed over at the end of the ten sessions; however, if you decide to abandon the experiment, you will be paid for the sessions you have attended to.

If you decide to take part in the experiments (and I hope you will), the first session will be devoted to discuss at length any query you could have about the procedures and, once you declare yourself fully satisfied, a number of measurements will be performed, in order to adequately adjust the experimental setting. The measurements involved are: weight, stature, height to your shoulders, full length of your arms, length of your forearms, breadth of your shoulders, and the height of your hands in the experimental posture.

For the session number 2 you will be asked to adopt the experimental posture (which will be defined in terms of your own dimensions) and hold it until you experience a level of discomfort such that it makes impossible to continue the effort.

It is crucial that you identify and remember as clearly as possible the level of discomfort that made you stop the effort during this session, since you will be asked to attempt a maximum holding during each of the remaining sessions and it is expected that you will stop when you reach exactly the same level of discomfort.

The rest of the sessions (3 to 10) will be designed around the longest time you were able to hold the posture in session 2, which is called maximum holding time (MHT). The pattern is that you will be asked to hold the posture for a proportion of that time, then you will be given a rest period and after this, you will hold the posture once again for as long as you can.

It is essential to learn how discomfort builds-up and which areas of your body are the most affected. To reach both these objectives we will combine the use of a scale that allows you to describe the sensations you are experiencing by assigning it a number, together with a diagram that shows those parts of your body which, according to findings from similar research in the past, are more likely to be affected. There is a separate set of instructions referring to the way I expect you to use both the scale and the diagram.

Instructions for the Estimation of Body Part Discomfort

A) In this experiment you are asked to hold a posture until you reach a degree of discomfort such that it prevents you from continuing the effort; ideally, this should be the most unpleasant sensation you can stand at that particular moment. Clearly, we are not aiming to see how far a person can be pushed into enduring an unpleasant situation and you should not feel compelled to "do better" each time.

B) It will help if you think of discomfort as those unpleasant physical sensations (such as tingling, warmth, throbbing, etc.) arising from the prolonged holding of a poor or demanding posture, sensations that could eventually develop into pain. You may get a good example of the sort of discomfort we are talking about by stretching your arm above your head and holding it there for as long as you can; you will see how quickly you start experiencing the sensations mentioned above.

C) During the experiment you will have to estimate and express the level of discomfort you are experiencing in a number of areas of your body, as shown in the diagram in front of you.

D) Once every minute, the experimenter will call the different areas and you will assign to each of them a number taken from the scale also shown in front of you.

E) When estimating the discomfort, concentrate on the muscles and joints contained in each area and choose the number that best describes the level of sensation you are experiencing in that area at that moment.

F) It will be better if you begin assigning the numbers in a rather conservative fashion, so as to avoid "running out of scale" at the top. As a rough guide, maximal represents for most of the people the point where they have to stop because they feel exhausted. This is the moment when you will call "stop".

G) It is normal that discomfort builds up more quickly in some areas; do not feel you have to assign numbers as close as possible for adjacent areas. Of course, it could even be the case that some areas do not exhibit any discomfort at all.

H) You could find that an area reaches a certain level of discomfort and then it remains the same or even decreases. Once again, this is perfectly normal and you must express what you feel at that particular moment.

This sheet with instructions for the estimation of body parts discomfort was handed to the subject in both experimental stages.

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

SCHEDULE OF EXPERIMENTAL SESSIONS

Dear ____, I have arranged for you to attend experimental sessions at the following dates and times:____

____

____

____

____

The sessions will take place at the Ergonomics Laboratory, located in the first floor of the New Laboratories Building, at the top of the left-hand side flight of stairs.

I hope you will find convenient the times I have assigned for your attendance. However, if you find it difficult to keep any of the appointments, I will be very grateful if you care to give me as advanced notice as possible, in order to re-schedule your own subsequent times and to minimise the disturbance to the rest of the experiments.

To facilitate the observations and measurements to be performed, it will be necessary that every time you come to the lab, you wear a short-sleeved shirt or blouse, without padding on the shoulders. I hope you will not find this an unreasonable request.

To prevent the appearance of any kind of prejudice that could later on affect the results of this experiment, may I ask you not to comment with any of your fellow subjects about the procedures you will be experiencing during the different sessions?

Once again, I thank you for your cooperation with this project. I look forwards to see you in the Lab.

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

Subject Consent Form

I understand that the purpose of this experiment is to measure the length of time for which a posture can be maintained and that the information thus obtained will be used to help in the design of workplaces. The experiment will be carried out in several sessions over a period of 4 weeks.

I have been given a description of the tests and measurements to be made. I realise that the tests may be fatiguing and that some discomfort could result from my participation.

I fully understand that I may at any time withdraw from taking part in the experiments. My replies to those questions concerning the state of my health and my fitness to participate in this study, which are attached to this consent form, are correct to the best of my knowledge.

I agree to the publication of the results of the experiments, on the understanding that these are in coded form and my identity cannot be inferred from them.

I hereby volunteer to participate as an experimental subject in the tests during the period ..../..................../1991 to ..../................../1991.

Signature: ....

Date: ....

WANTED

I AM CONDUCTING AN ERGONOMICS EXPERIMENT AND NEED FEMALE SUBJECTS AGED 18 TO 24.

MY RESEARCH LOOKS AT HOW LONG PEOPLE CAN HOLD A WORKING POSTURE.

YOU WILL ATTEND 10 SESSIONS (AROUND 30 MINS EACH) OVER 4 WEEKS.

YOU WILL BE RICHER AT THE END OF THE EXPERIMENT (£15!!!).

IF INTERESTED, PLEASE RING INTERNAL 3807 AND ASK FOR MR. NIEVES SERRATOS.

THANK YOU.

Copy of the poster used to invite volunteers to take part in the first experimental stage of the study.

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

Information Package

INSTITUTE FOR OCCUPATIONAL ERGONOMICS

DEPARTMENT OF MANUFACTURING ENGINEERING
AND OPERATIONS MANAGEMENT

UNIVERSITY OF NOTTINGHAM

October 1992

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

Basic Information

This research project is located in the realm of occupational ergonomics; its ultimate goal is a contribution to improve the conditions in which an important number of workers perform their jobs. The aim of the experiments we are about to undertake is to obtain a better understanding of the way fatigue develops when a person adopts a series of postures that have been observed to occur in actual work situations.

The experimental postures involve standing with both upper arms partially raised to the sides (at angles of 30, 60 and 90 degrees), in line with your body; elbows flexed in a right angle; forearms parallel to the floor; each hand touching a small piece of a plastic sheet located in front of you. The whole experiment consists of 10 sessions and in all but the first of them you will be asked to adopt one of the described postures and hold it for as long as you can, until you have to stop through fatigue. This will certainly involve a certain degree of discomfort, but this should not last for long after the session has finished. Any two consecutive sessions will be separated by at least 48 hours, to ensure that you are fully recovered.

IT IS NOT THE AIM OF THE EXPERIMENT TO SUBMIT YOU TO UNDUE FATIGUE, BUT ONLY TO KNOW HOW LONG IT TAKES FOR THE POSTURES TO BECOME UNBEARABLE. ALL I WILL BE ASKING OF YOU IS TO BE PREPARED TO MAKE AN HONEST EFFORT IN EVERY SESSION AND HOLD THE POSTURE TO YOUR TRUE LIMIT.

Each experimental session will last around 120 minutes and the times may be arranged to suit your needs. You will be paid at a rate of £5.00 per session; *the money will be handed over at the end of the ten sessions.*

The degree of discomfort you are experiencing will be monitored in two different ways. One of these will require you to rate the discomfort according to how strong you perceive it to be; there is a separate set of instructions for this particular issue and they are included with this information package.

The other way of monitoring the development of discomfort will be by recording the level of electrical activity generated by those muscles most involved in the effort of holding the posture being studied, namely, the muscles around your shoulders. The procedure, called electromyography, is very similar to that involved in taking an electrocardiogram, something that most probably you have seen in movies or T.V. programmes such as "Casualty" and the like.

Since electromyography requires the attachment of electrodes directly onto the skin; if you are a male, you will be asked to remove all your clothes from above the waist; if you are a female, you will be asked to wear a sleeveless top that leaves uncovered the shoulders and the back immediately below the neck.

In the first experimental session a number of measurements will be performed, in order to adequately adjust the experimental setting. The measurements involved are: weight, stature, height to your shoulders, full length of your arms, length of your forearms, and the height and distance between your hands in the experimental postures. Additionally, I will need to record the response of your muscles when you are asked to hold in your hand a known load, such that I can "calibrate" against this the responses I will be getting in the rest of the sessions.

Please, feel free to enquire about anything to do with the experimental procedure at any time you require to do so, for I will do my best to offer you a proper answer. It is my utmost interest that you are fully satisfied and appreciating (hopefully enjoying) the experience of taking part in my experiments.

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

SCHEDULE OF EXPERIMENTAL SESSIONS

Dear ____, I have arranged for you to attend experimental sessions at the following dates and times:____

____

____

____

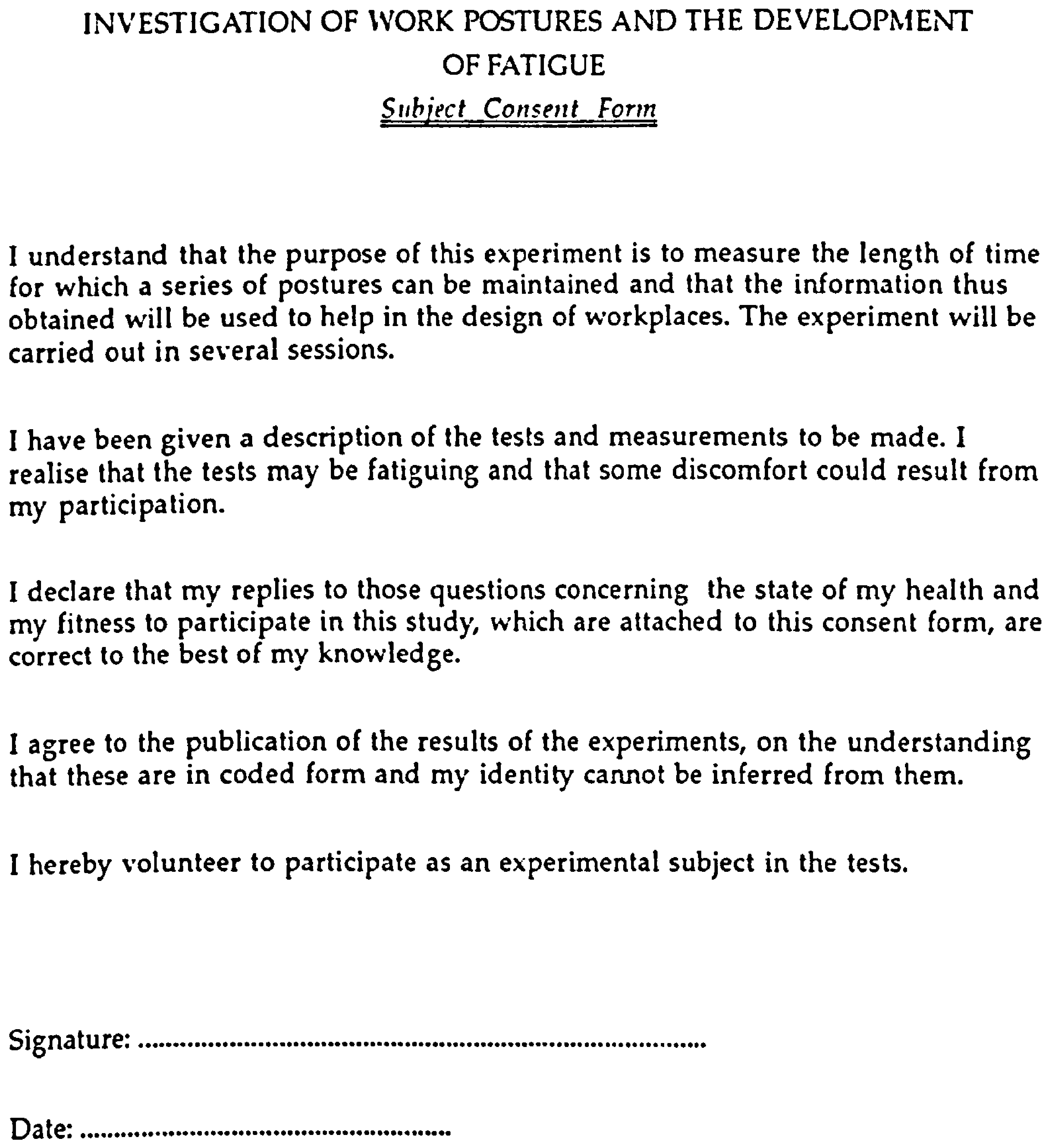
The sessions will take place at the Ergonomics Laboratory, located in the first floor of the New Laboratories Building, at the top of the left-hand side flight of stairs.

I hope you will find convenient the times I have assigned for your attendance. However, if you find it difficult to keep any of the appointments, I will be very grateful if you care to give me as advanced notice as possible, in order to re-schedule your own subsequent times and to minimise the disturbance to the rest of the experiments.

If you are a female subject, let me remind you that it will be necessary that every time you come to the lab, you wear a sleeveless top that leaves uncovered the shoulders and the back immediately below the neck.

To prevent the appearance of any kind of prejudice that could later on affect the results of this experiment, may I ask you not to comment with any of your fellow subjects about the procedures you will be experiencing during the different sessions?Once again, I thank you for your cooperation with this project. I look forwards to see you in the Lab.

Sesion scheduler prepared for the second stage of the study.



INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

Subject Consent Form

I understand that the purpose of this experiment is to measure the length of time for which a series of postures can be maintained and that the information thus obtained will be used to help in the design of workplaces. The experiment will be carried out in several sessions.

I have been given a description of the tests and measurements to be made. I realise that the tests may be fatiguing and that some discomfort could result from my participation.

I declare that my replies to those questions concerning the state of my health and my fitness to participate in this study, which are attached to this consent form, are correct to the best of my knowledge.

I agree to the publication of the results of the experiments, on the understanding that these are in coded form and my identity cannot be inferred from them.

I hereby volunteer to participate as an experimental subject in the tests.

Signature: ....

Date: ....

Consent form used in the second experimental stage

WANTED

MALE AND FEMALE SUBJECTS, AGED 18 TO 24, TO TAKE PART IN ERGONOMICS EXPERIMENT.

THE RESEARCH LOOKS AT HOW LONG PEOPLE ARE ABLE TO HOLD A WORKING POSTURE.

YOU WILL NEED TO ATTEND TO 10 SESSIONS (90 - 120 MINUTES EACH) OVER 5-6 WEEKS.

YOU WILL BE PAID £5 PER SESSION. MONEY TO BE HANDED AS A LUMP SUM AT THE END OF THE EXPERIMENT.

IF INTERESTED, PLEASE RING MR. NIEVES SERRATOS, ON INTERNAL 4036/8083; OR COME AND SEE ME AT ROOM C-3, IOE BUILDING, OR AT THE ERGONOMICS LAB, LABS BLOCK.

THANK YOU.

13/10/92

Copy of the poster used to invite volunteers to take part in the second series of trials. (Notice the increase of the financial reward on offer).

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

Preliminary Questionnaire

1. Have you suffered from any serious illness over the last six months?

YES/NO

2. At present, are you having trouble of any sort with joints or muscles?

YES/NO

3. Do you have a job?

YES/NO

If so, does your job involve a heavy physical exertion?

YES/NO

Do you have to spend long time in a fixed posture?

YES/NO

4. Are you currently involved in any of the following sport/leisure activities? If so, please state frequency and/or intensity.

	How often?	Do you feel tired more than 1 day afterwards?
Tennis		YES/NO
Squash		YES/NO
Swimming		YES/NO
Weight lifting		YES/NO
Badminton		YES/NO

This preliminary questionnaire was asked verbally by the researcher when the potential subjects expressed their interest in the experiment.

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE

Health/Fitness Questionnaire

Name: ....

Age: ....

Address:

1. Have you ever suffered from a serious illness?
YES/NO
If "yes", please give brief particulars and approximate date

2. Have you ever been injured seriously, i.e., badly enough as to be treated by a
doctor or taken to a hospital?
YES/NO
If "yes", please give brief particulars and approximate date:

3. Are you at present under medical treatment of any kind?
YES/NO
If "yes", please indicate what kind of treatment (e.g. medicines, appliances,
physiotherapy, dressings)

4. Do you suffer from any disability which affects your daily life, work or
travelling?
YES/NO
If "yes", please give brief particulars:

5. Do you suffer from or have you in the past suffered from, any of the following
conditions:

a) Back pain or back problems;

b) Neck or shoulder strain;

c) Heart trouble;

d) Diabetes;

e) Hernia;

f) Chronic headaches;

g) Hypertension?

If "yes", please give brief particulars

Health questionnaire completed by the subjects themselves. The format has been slightly modified so to conform with that of the thesis.

INVESTIGATION OF WORK POSTURES AND THE DEVELOPMENT OF FATIGUE
DATA FROM DISCOMFORT RATING

STAGE.....   DATE...../......./......./   SUBJECT No...........   SESSION No......   REGIME..................

TIME	A	B	C	D	E		F		G		H		J		K		L	
					1	2	1	2	1	2	1	2	1	2	1	2	1	2

Copy of the form used in the collection of the discomfort ratings.

Appendix B

Relevant statistics of the regression lines fitted to the data of discomfort rating and holding time collected during each of the 89 trials

Subject	Trial No.	Abduction angle (deg)	Correlation coefficient between discomfort rating and % MHT	Slope of the regression line for the complete set of data	Slope of the regression line for the reduced set of data
1	1	30	0.9110	0.088	0.083
1	2	30	0.9849	0.109	0.113
1	3	30	0.9849	0.104	0.106
1	1	60	0.9798	0.108	0.112
1	2	60	0.9899	0.107	0.111
1	3	60	0.9747	0.112	0.112
1	1	90	0.7874	0.113	0.134
1	2	90	0.9747	0.106	0.116
1	3	90	0.9899	0.108	0.115
2	1	30	0.9798	0.091	0.090
2	2	30	0.9798	0.095	0.094
2	3	30	0.9644	0.090	0.088
2	1	60	0.9899	0.102	0.102
2	2	60	0.9695	0.108	0.110
2	3	60	0.9274	0.085	0.081
2	1	90	0.9644	0.102	0.103
2	2	90	0.9381	0.084	0.086
2	3	90	0.9644	0.095	0.092

Subject	Trial No.	Abduction angle (deg)	Correlation coefficient between discomfort rating and % MHT	Slope of the regression line for the complete set of data	Slope of the regression line for the reduced set of data
3	1	30	0.9899	0.107	0.108
3	2	30	0.9539	0.121	0.123
3	3	30	0.9487	0.121	0.123
3	1	60	0.9539	0.118	0.123
3	2	60	0.9644	0.122	0.126
3	3	60	0.9487	0.123	0.128
3	1	90	0.7810	0.123	0.132
3	2	90	0.9849	0.108	0.110
3	3	90	0.9220	0.117	0.123
4	1	30	0.9899	0.107	0.109
4	2	30	0.9644	0.095	0.093
4	3	30	0.9644	0.103	0.104
4	1	60	0.9899	0.103	0.105
4	2	60	0.9798	0.100	0.100
4	3	60	0.9899	0.104	0.105
4	1	90	0.9950	0.095	0.089
4	2	90	0.9798	0.112	0.112
4	3	90	0.9381	0.099	0.099

Subject	Trial No.	Abduction angle (deg)	Correlation coefficient between discomfort rating and % MHT	Slope of the regression line for the complete set of data	Slope of the regression line for the reduced set of data
5	1	30	0.9798	0.094	0.092
5	2	30	0.9644	0.093	0.092
5	3	30	0.9695	0.099	0.099
5	1	60	0.9000	0.072	0.099
5	2	60	0.9798	0.091	0.088
5	3	60	0.9798	0.092	0.090
5	1	90	0.9899	0.095	0.091
5	2	90	0.8888	0.072	0.089
5	3	90	0.9539	0.092	0.088
6	1	30	0.9327	0.080	0.088
6	2	30	0.9539	0.083	0.081
6	3	30	-----	-----	-----
6	1	60	0.9899	0.098	0.098
6	2	60	0.9747	0.095	0.095
6	3	60	0.9899	0.102	0.102
6	1	90	0.9695	0.090	0.086
6	2	90	0.9899	0.096	0.094
6	3	90	0.9798	0.100	0.101

Subject	Trial No.	Abduction angle (deg)	Correlation coefficient between discomfort rating and % MHT	Slope of the regression line for the complete set of data	Slope of the regression line for the reduced set of data
7	1	30	0.9747	0.094	0.093
7	2	30	0.9899	0.106	0.107
7	3	30	0.9695	0.096	0.095
7	1	60	0.9695	0.092	0.090
7	2	60	0.9487	0.08	0.093
7	3	60	0.9695	0.090	0.087
7	1	90	0.9540	0.088	0.081
7	2	90	0.9644	0.091	0.087
7	3	90	0.9849	0.098	0.097
8	1	30	0.9798	0.109	0.111
8	2	30	0.9110	0.077	0.092
8	3	30	0.9798	0.101	0.101
8	1	60	0.9695	0.088	0.083
8	2	60	0.8944	0.083	0.088
8	3	60	0.9798	0.098	0.097
8	1	90	0.9274	0.093	0.088
8	2	90	0.9592	0.087	0.087
8	3	90	0.9899	0.090	0.086

Subject	Trial No.	Abduction angle (deg)	Correlation coefficient between discomfort rating and % MHT	Slope of the regression line for the complete set of data	Slope of the regression line for the reduced set of data
9	1	30	0.9849	0.100	0.099
9	2	30	0.9849	0.100	0.100
9	3	30	0.9950	0.101	0.101
9	1	60	0.9747	0.100	0.100
9	2	60	0.9747	0.097	0.096
9	3	60	0.9798	0.101	0.101
9	1	90	0.9798	0.094	0.093
9	2	90	0.9695	0.097	0.097
9	3	90	0.9695	0.092	0.090
10	1	30	0.9950	0.107	0.107
10	2	30	0.9849	0.087	0.095
10	3	30	0.9950	0.102	0.102
10	1	60	0.9849	0.111	0.112
10	2	60	0.9849	0.100	0.101
10	3	60	0.9695	0.118	0.120
10	1	90	0.9747	0.105	0.106
10	2	90	0.9747	0.096	0.095
10	3	90	0.9695	0.093	0.091

Appendix C

Percentage change in RMS amplitude and MPF observed during each of the 89 trials

NOTE: Percentage changes in RMS amplitude are shown in pages 418-422;
Percentage changes in MPF are shown in pages 423-427.

Details of the trial			Percentage change in RMS amplitude from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
1	1	30	0	0	2	12	-7	-6
1	2	30	2	3	0	17	17	9
1	3	30	2	2	0	6	19	0
1	1	60	4	0	2	8	-2	5
1	2	60	0	-5	0	3	-13	0
1	3	60	6	6	2	11	35	17
1	1	90	2	-18	0	0	2	-4
1	2	90	2	-2	2	5	-4	-7
1	3	90	0	0	2	20	-20	13
2	1	30	3	22	2	110	95	21
2	2	30	6	16	4	6	26	36
2	3	30	11	5	4	6	14	9
2	1	60	13	27	6	117	102	70
2	2	60	23	11	14	45	----	203
2	3	60	0	5	4	12	16	32
2	1	90	0	2	2	136	18	32
2	2	90	2	33	4	35	88	43
2	3	90	6	16	15	11	44	9

Details of the trial			Percentage change in RMS amplitude from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
3	1	30	12	61	----	109	29	5
3	2	30	8	23	0	39	6	9
3	3	30	13	22	2	37	9	9
3	1	60	6	17	0	18	7	13
3	2	60	7	7	0	41	10	9
3	3	60	9	14	2	26	21	14
3	1	90	19	5	4	65	-2	22
3	2	90	----	----	-6	----	----	-8
3	3	90	6	21	21	57	50	23
4	1	30	65	73	83	53	-6	30
4	2	30	86	25	47	35	-15	16
4	3	30	43	61	44	23	-11	23
4	1	60	20	-13	33	21	-2	50
4	2	60	35	-12	59	71	-13	46
4	3	60	60	53	129	64	-34	----
4	1	90	8	17	29	4	28	28
4	2	90	49	20	68	32	13	54
4	3	90	59	25	100	74	9	80

Details of the trial			Percentage change in RMS amplitude from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
5	1	30	54	-21	13	307	-2	-31
5	2	30	68	-12	42	273	9	13
5	3	30	124	15	98	144	19	35
5	1	60	47	-18	15	44	-6	9
5	2	60	31	-25	14	117	-28	29
5	3	60	27	-33	15	106	-11	54
5	1	90	50	-38	6	105	-19	-11
5	2	90	16	-7	20	52	-14	44
5	3	90	36	0	58	60	-3	157
6	1	30	33	35	0	32	21	-12
6	2	30	82	17	13	89	19	14
6	3	30	----	----	----	----	----	----
6	1	60	85	102	2	35	61	13
6	2	60	37	29	6	50	32	9
6	3	60	27	5	4	92	27	9
6	1	90	40	2	8	73	14	17
6	2	90	30	-5	8	42	16	13
6	3	90	29	-6	2	45	29	12

Details of the trial			Percentage change in RMS amplitude from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
7	1	30	11	19	0	15	12	4
7	2	30	29	10	0	39	9	0
7	3	30	34	7	0	48	3	0
7	1	60	24	20	13	27	31	8
7	2	60	55	24	17	85	34	9
7	3	60	54	28	4	76	30	14
7	1	90	19	60	8	19	52	13
7	2	90	49	2	30	40	33	13
7	3	90	20	-13	31	42	43	44
8	1	30	2	-38	2	-2	-39	0
8	2	30	11	2	2	44	-10	4
8	3	30	-1	8	-6	29	12	9
8	1	60	12	-18	2	20	-22	8
8	2	60	14	-27	0	33	-5	4
8	3	60	10	-10	14	76	3	28
8	1	90	33	2	6	29	3	15
8	2	90	6	-8	12	74	16	36
8	3	90	53	22	23	81	-4	41

Details of the trial			Percentage change in RMS amplitude from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
9	1	30	107	178	95	215	208	74
9	2	30	176	86	138	144	121	126
9	3	30	140	38	11	59	89	54
9	1	60	77	79	167	215	204	150
9	2	60	137	51	121	179	154	95
9	3	60	52	139	281	100	171	171
9	1	90	----	28	196	208	182	250
9	2	90	65	89	191	143	128	178
9	3	90	308	26	252	170	196	485
10	1	30	12	156	200	55	165	36
10	2	30	-10	163	133	63	104	79
10	3	30	30	108	200	-14	156	21
10	1	60	17	62	300	150	50	180
10	2	60	48	49	455	102	139	250
10	3	60	21	67	230	107	82	279
10	1	90	159	7	182	157	18	44
10	2	90	150	76	310	73	51	268
10	3	90	37	48	220	60	109	167

Details of the trial			Percentage change in MPF from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
1	1	30	-3	-5	-6	0	0	0
1	2	30	-2	-7	1	-2	-4	0
1	3	30	2	-6	-9	12	-1	0
1	1	60	3	-15	-13	9	-2	7
1	2	60	-14	-15	-9	7	-10	5
1	3	60	10	-19	-5	0	-13	-6
1	1	90	9	-15	-4	7	-4	3
1	2	90	-6	-18	0	8	-21	-3
1	3	90	4	-22	-18	2	-29	-3
2	1	30	1	-10	-3	-4	-19	0
2	2	30	18	-11	-22	2	-34	-25
2	3	30	-2	-21	-21	12	-16	20
2	1	60	9	-19	-23	-16	-46	-28
2	2	60	11	-17	-18	31	----	-42
2	3	60	12	-16	-16	26	-34	-14
2	1	90	12	-28	-11	-16	-40	-10
2	2	90	6	-36	-36	-8	-29	-28
2	3	90	7	-38	-28	-5	-33	-15

Details of the trial			Percentage change in MPF from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
3	1	30	-19	-14	----	-11	-10	-16
3	2	30	-7	-7	25	-11	-10	-18
3	3	30	-3	-6	-21	-11	-16	-22
3	1	60	1	-18	-15	8	-10	-8
3	2	60	5	-20	-13	0	-13	-17
3	3	60	-4	-16	-10	5	-14	-9
3	1	90	-7	-27	-26	5	-21	-26
3	2	90	----	-18	----	----	-34	-5
3	3	90	-6	-19	-23	-7	-16	-36
4	1	30	-2	-11	-3	5	-8	-4
4	2	30	-5	-12	-22	-6	-12	-17
4	3	30	9	-8	-20	2	-16	-12
4	1	60	5	-9	-10	0	-18	-5
4	2	60	7	-13	-18	-5	-23	-17
4	3	60	5	-12	0	5	-16	----
4	1	90	5	-15	-6	-4	-23	0
4	2	90	0	-18	-14	3	-33	-13
4	3	90	10	-22	-6	2	-35	-24

Details of the trial			Percentage change in MPF from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
5	1	30	14	-14	-16	11	0	-6
5	2	30	12	-12	-9	15	0	-3
5	3	30	-5	-13	-23	-18	-6	-10
5	1	60	4	-14	-14	2	-8	-14
5	2	60	14	-9	-19	22	-3	-5
5	3	60	18	-4	-10	16	-8	-12
5	1	90	8	-14	-8	13	-16	-10
5	2	90	6	-21	-12	-10	-22	-12
5	3	90	-5	-28	-19	4	-23	-15
6	1	30	-14	-5	-4	-1	-17	-16
6	2	30	3	-9	2	0	-3	10
6	3	30	----	----	----	----	----	----
6	1	60	2	-10	-7	-8	-5	-4
6	2	60	4	4	8	0	1	3
6	3	60	-4	-16	-12	-3	-4	-3
6	1	90	-10	-23	-11	-10	-6	0
6	2	90	-3	-35	-4	-14	-13	-5
6	3	90	-10	-23	-27	-9	-10	-9

Details of the trial			Percentage change in MPF from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
7	1	30	-2	-10	-11	3	-8	2
7	2	30	-3	-11	-8	-2	0	4
7	3	30	0	-11	-18	-2	-4	0
7	1	60	3	-13	-8	6	-13	-2
7	2	60	-4	-13	-7	3	-4	-8
7	3	60	-3	-15	-17	5	-3	3
7	1	90	-17	-24	----	-21	-15	-24
7	2	90	-8	-38	-14	-6	-10	-1
7	3	90	-19	-36	-25	-11	-13	-14
8	1	30	-5	-16	-8	2	-17	-8
8	2	30	-13	-19	-11	-9	-10	-7
8	3	30	-10	-24	-6	-11	-23	-14
8	1	60	-8	-14	-10	-10	-23	6
8	2	60	-7	-15	-6	-7	-16	-9
8	3	60	-19	-17	-20	-12	-24	-6
8	1	90	-12	-24	-18	-10	-20	-9
8	2	90	0	-24	-18	-12	-23	-7
8	3	90	-16	-25	-23	-13	-22	-16

Details of the trial			Percentage change in MPF from					
			Right arm			Left arm		
Subject number	Order	Angle (deg)	Trapezius	Medial deltoid	Post. deltoid	Trapezius	Medial deltoid	Post. deltoid
9	1	30	3	-8	7	2	-5	4
9	2	30	-4	-14	-11	7	-5	3
9	3	30	-9	-14	-12	-4	-4	-11
9	1	60	22	-18	-13	13	8	-13
9	2	60	44	-21	-12	14	-4	-12
9	3	60	14	-18	4	-3	-1	23
9	1	90	----	-30	-13	9	-9	2
9	2	90	-14	-31	-17	-11	-1	-21
9	3	90	1	-36	-16	-3	-19	-13
10	1	30	-4	1	-27	6	0	-15
10	2	30	-5	4	-13	-18	1	-18
10	3	30	-29	11	-20	-17	1	-18
10	1	60	18	-3	-33	9	3	-20
10	2	60	10	-1	-8	-3	-8	-22
10	3	60	0	1	-5	-8	-10	-18
10	1	90	14	-14	0	35	-25	-6
10	2	90	8	-23	-9	-4	-38	-26
10	3	90	-12	-21	-8	-20	-22	-5

Appendix D

Representative plots of the changes in MPF and RMS amplitude

Guide to the contents of Appendix D:

Pages

Plots of the changes in RMS amplitude at 30° (figs D1 - D10) 429 - 438

Plots of the changes in RMS amplitude at 60° (figs D11 - D20) 439 - 448

Plots of the changes in RMS amplitude at 90° (figs D21 - D30) 449 - 458

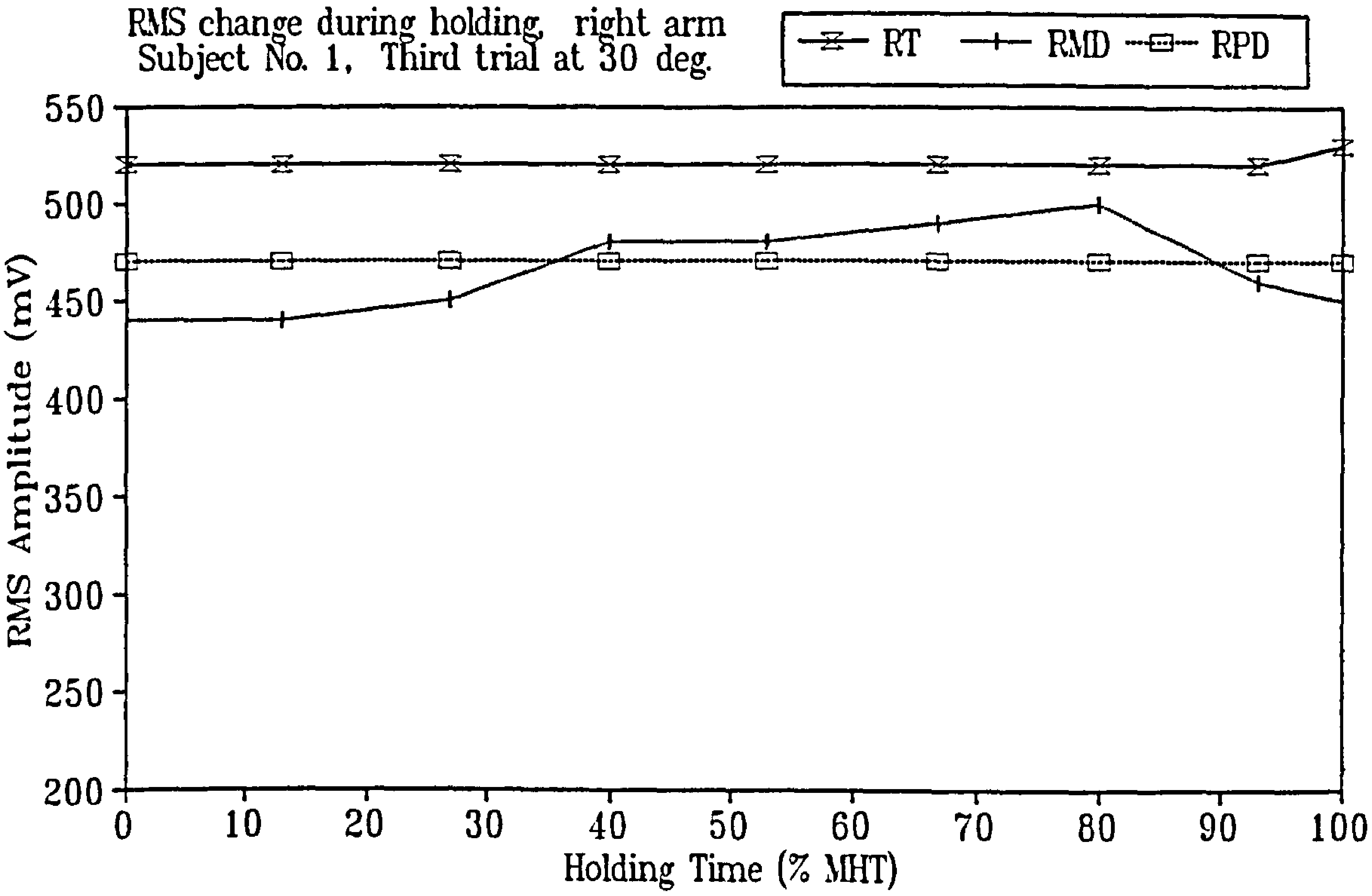
Plots of the changes in MPF at 30° (figs D31 - D40) 459 - 468

Plots of the changes in MPF at 60° (figs D41 - D50) 469 - 478

Plots of the changes in MPF at 90° (figs D51 - D60) 479 - 488

NOTES: 1) The trials illustrated are those with the largest value of correlation coefficient between holding time and the changes in EMG signal.

2) The changes that exhibited a significant reversed trend are marked with '*' in the key shown with each graph.



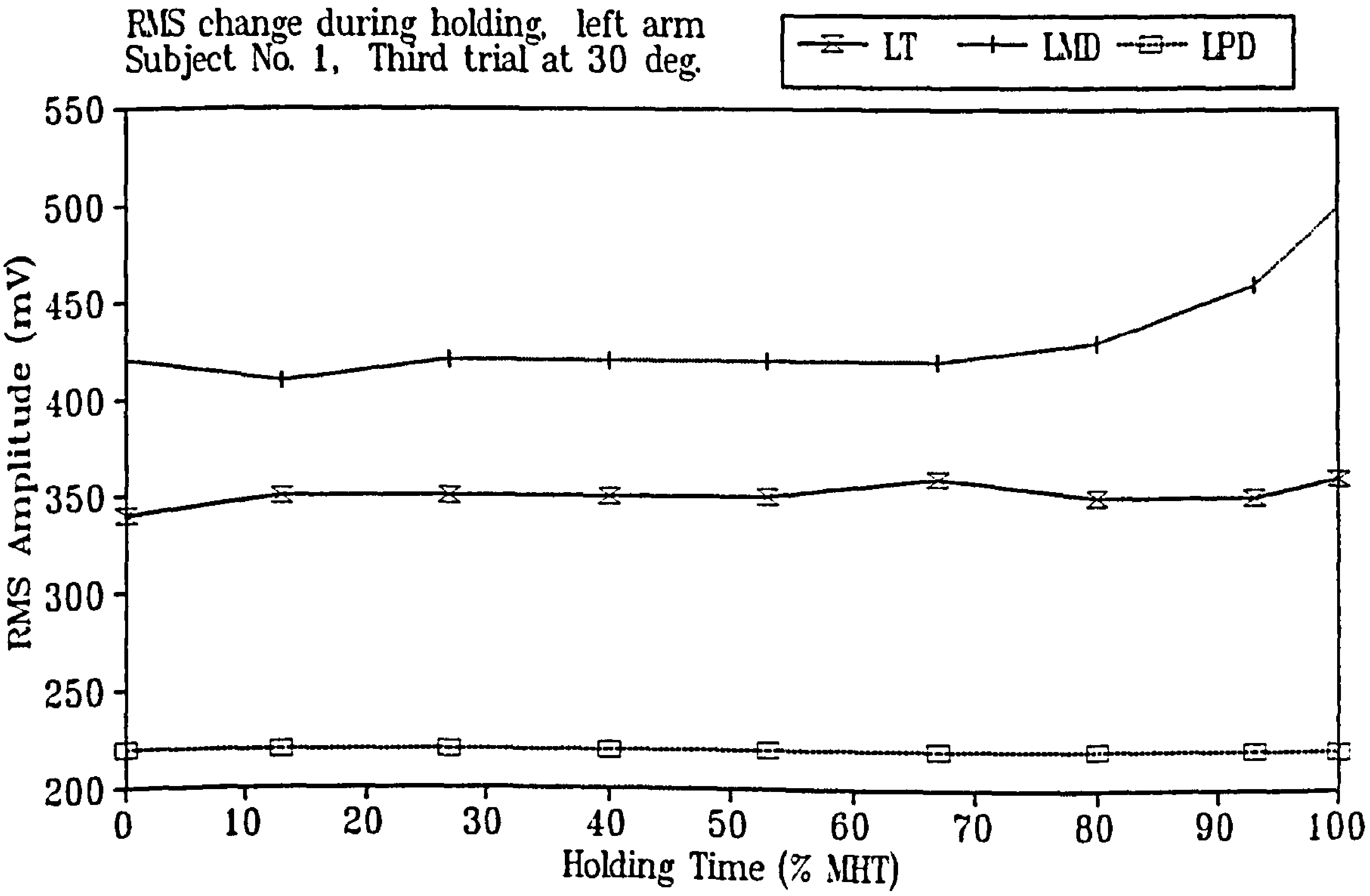
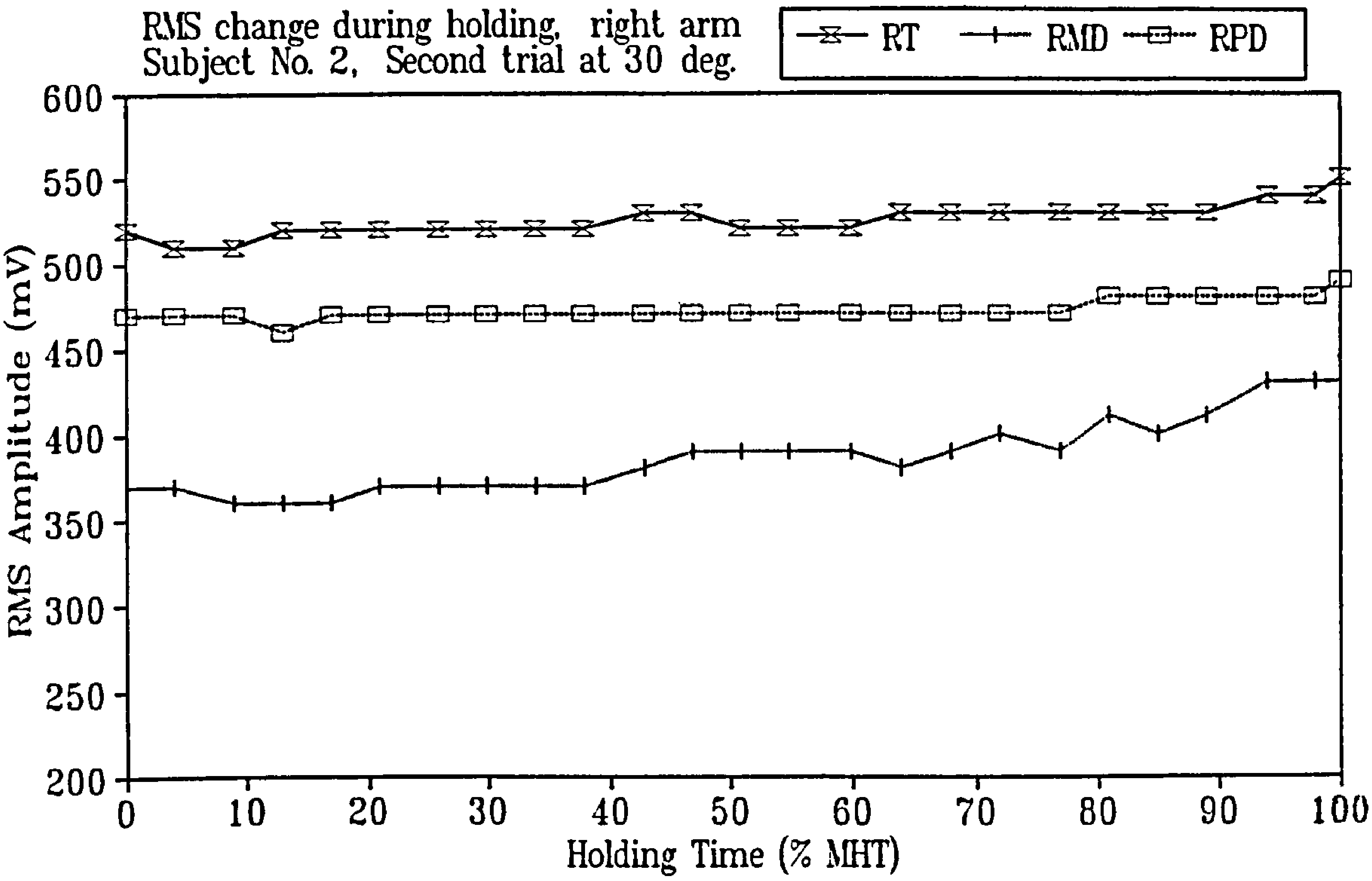


Figure D.1 Change in RMS amplitude for subject No. 1 at 30 deg.



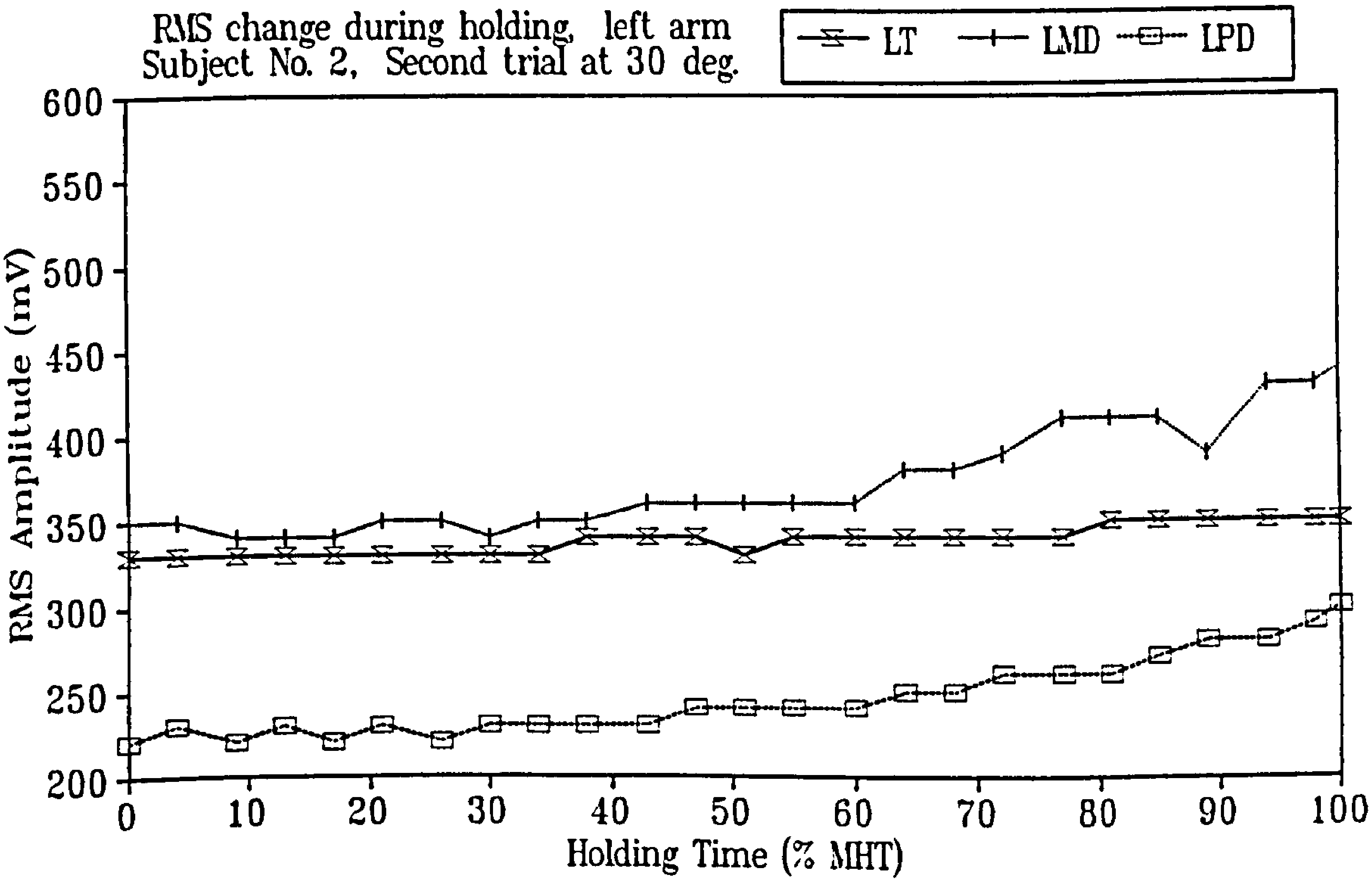
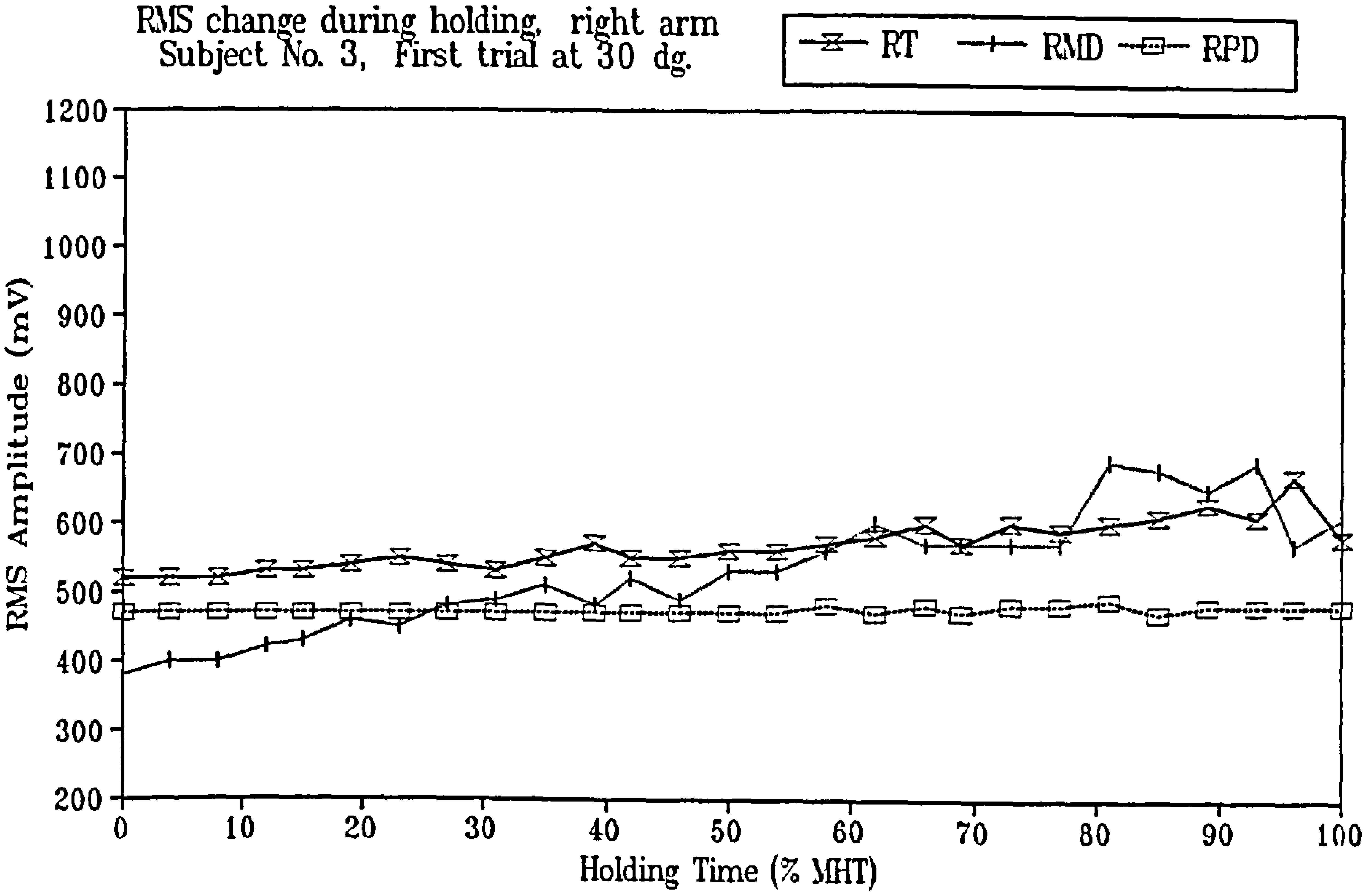


Figure D.2 Changes in RMS amplitude for subject No. 2 at 30 deg.



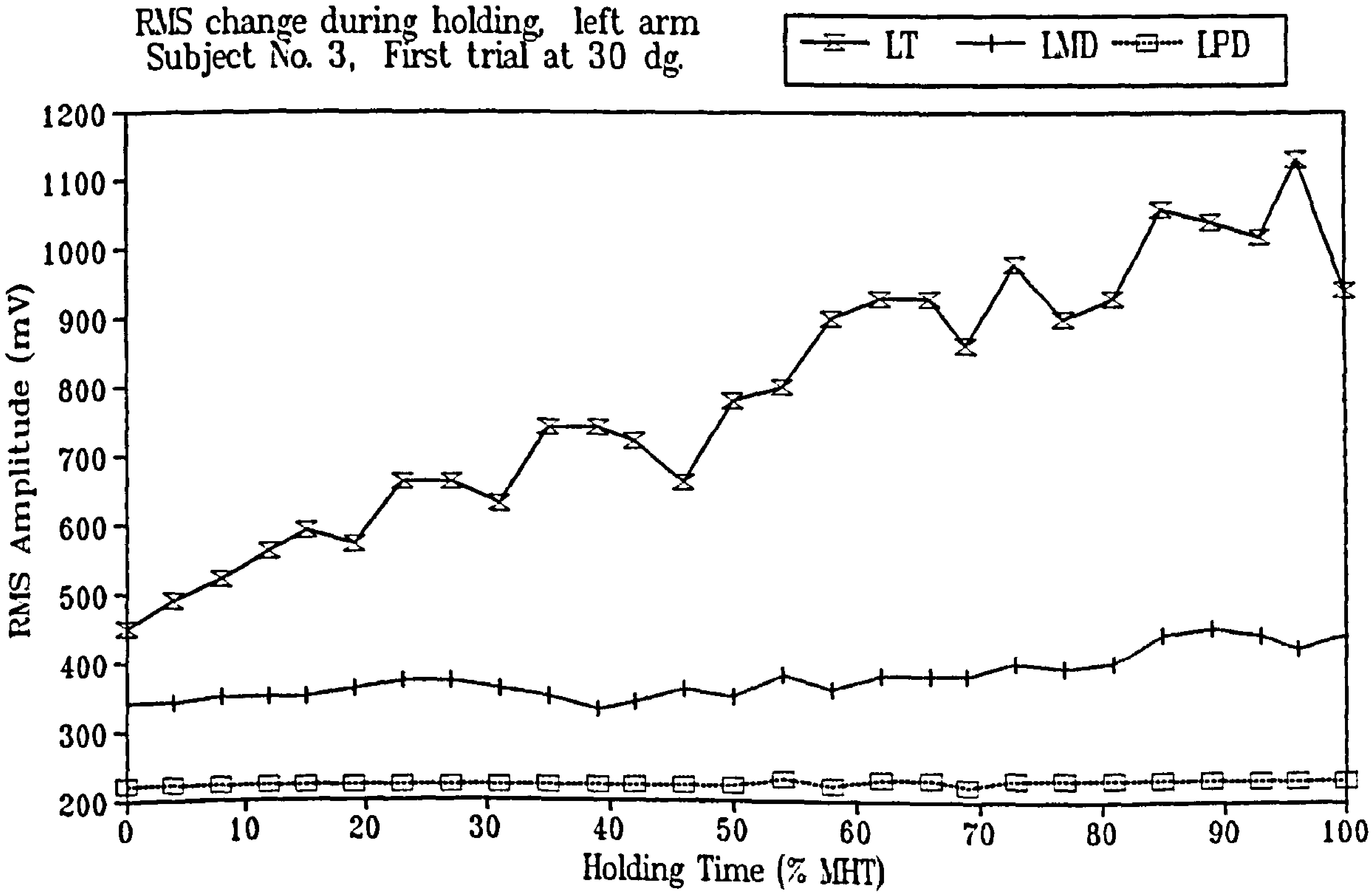
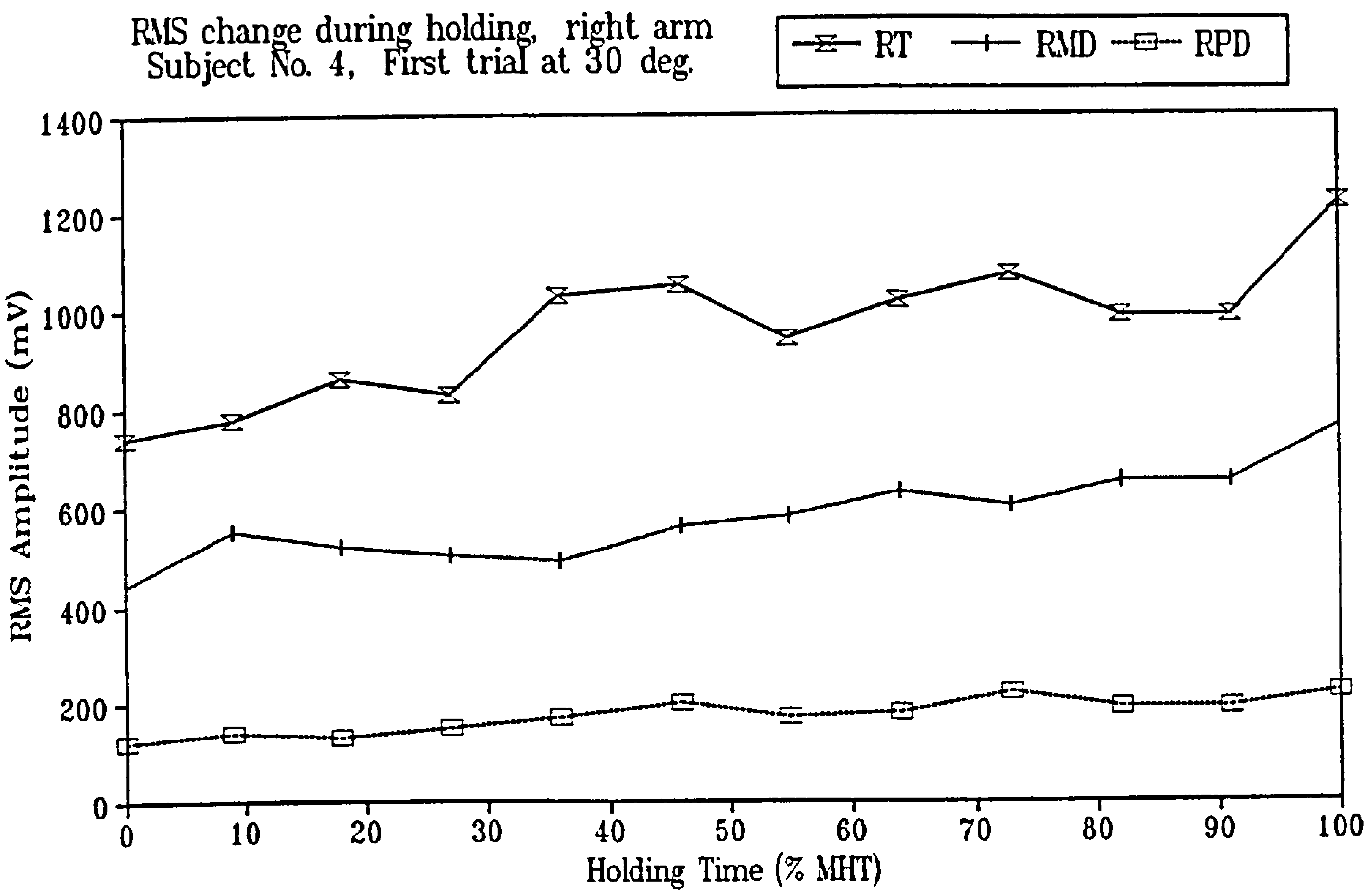


Figure D.3 Changes in RMS amplitude for subject No. 3 at 30 dg.



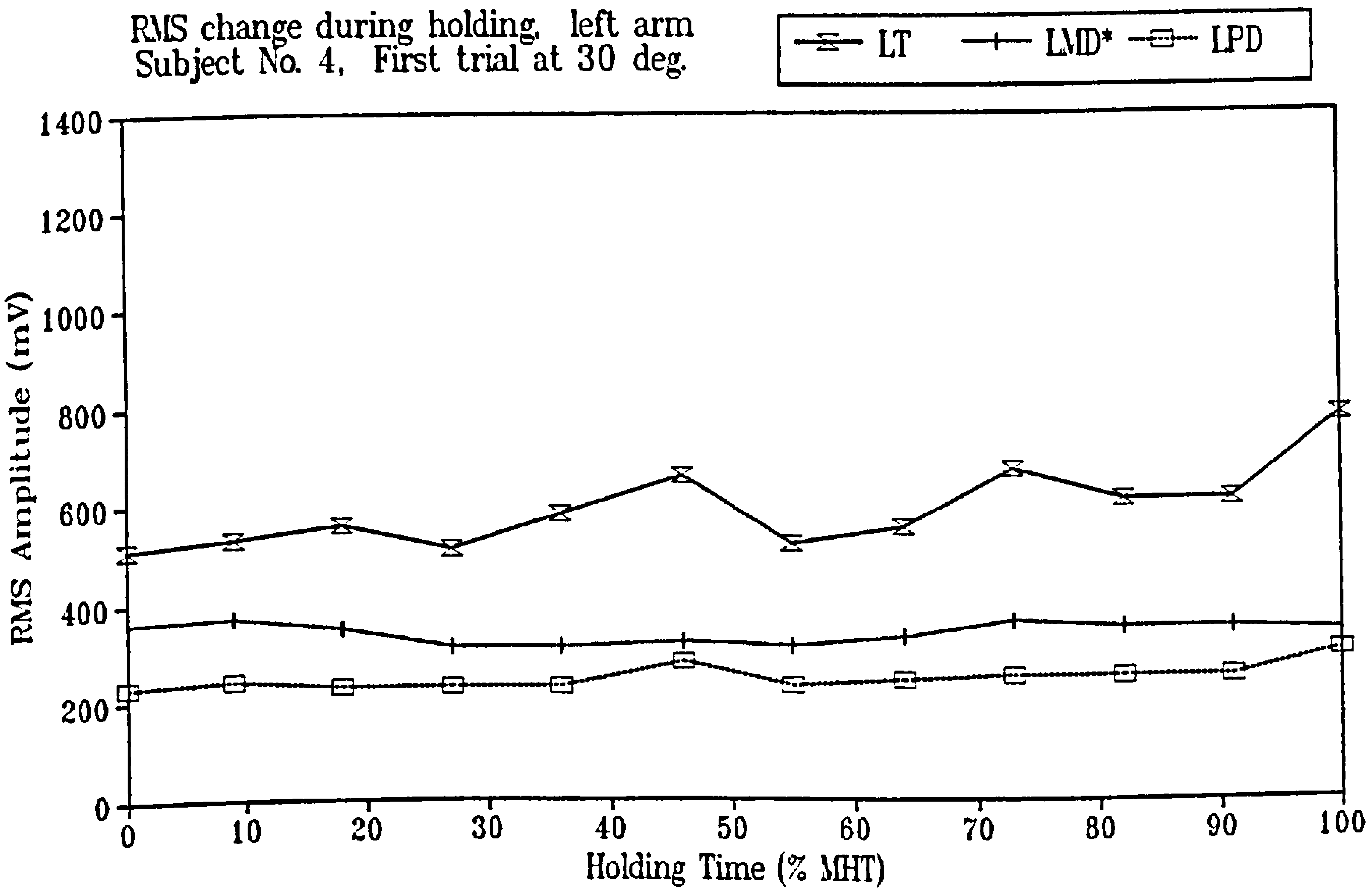
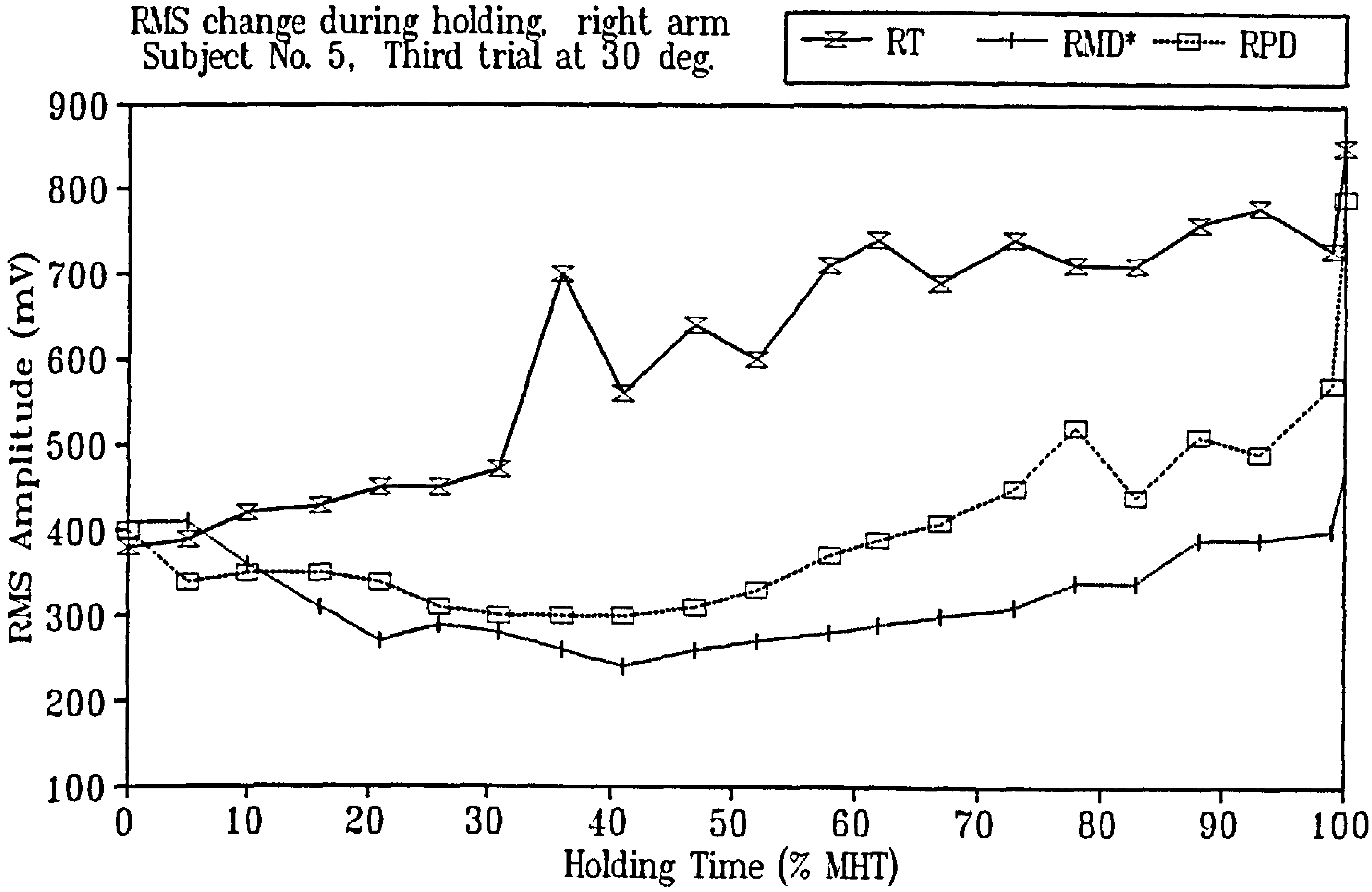


Figure D.4 Changes in RMS amplitude for subject No. 4 at 30 deg.



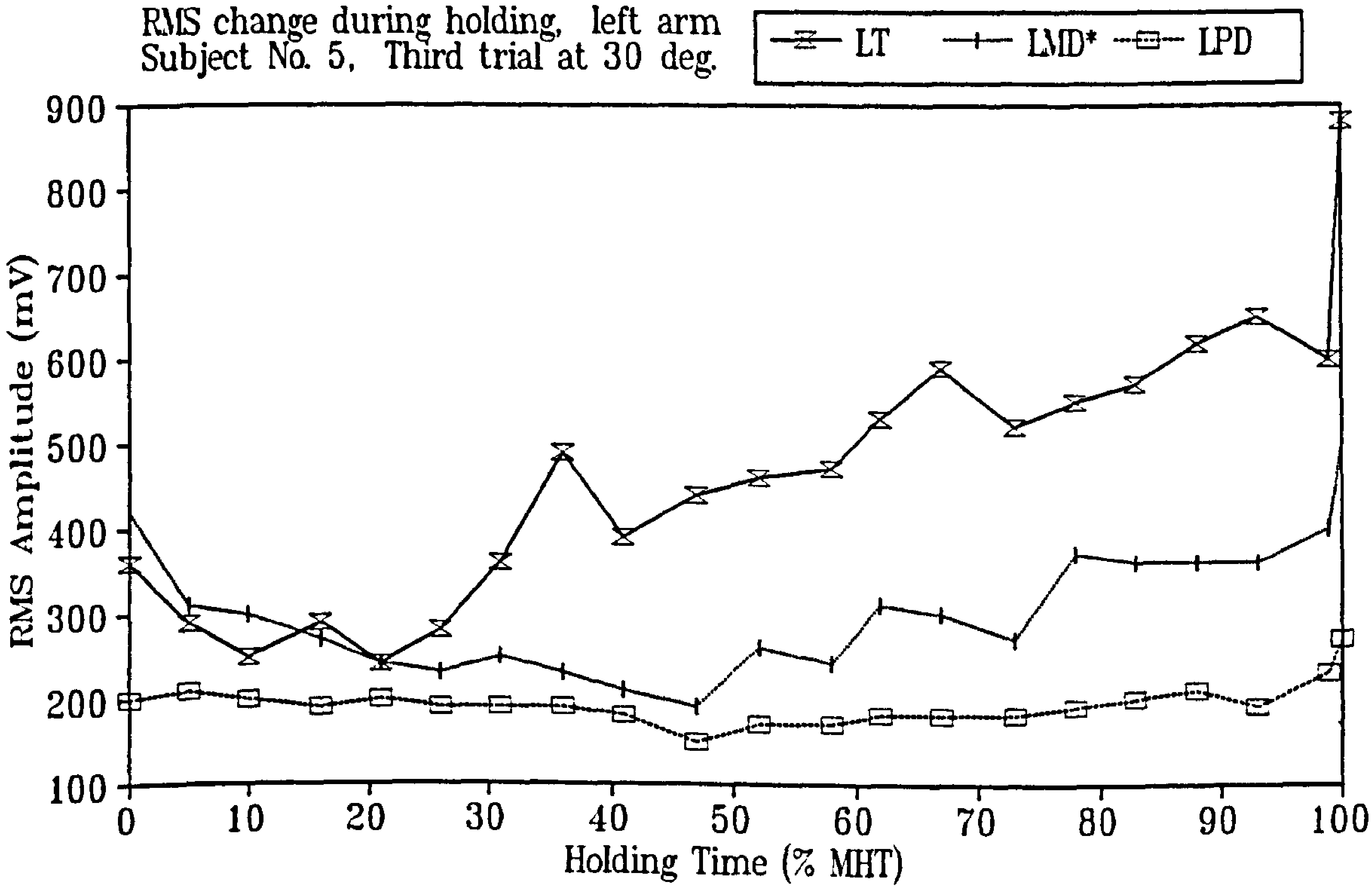
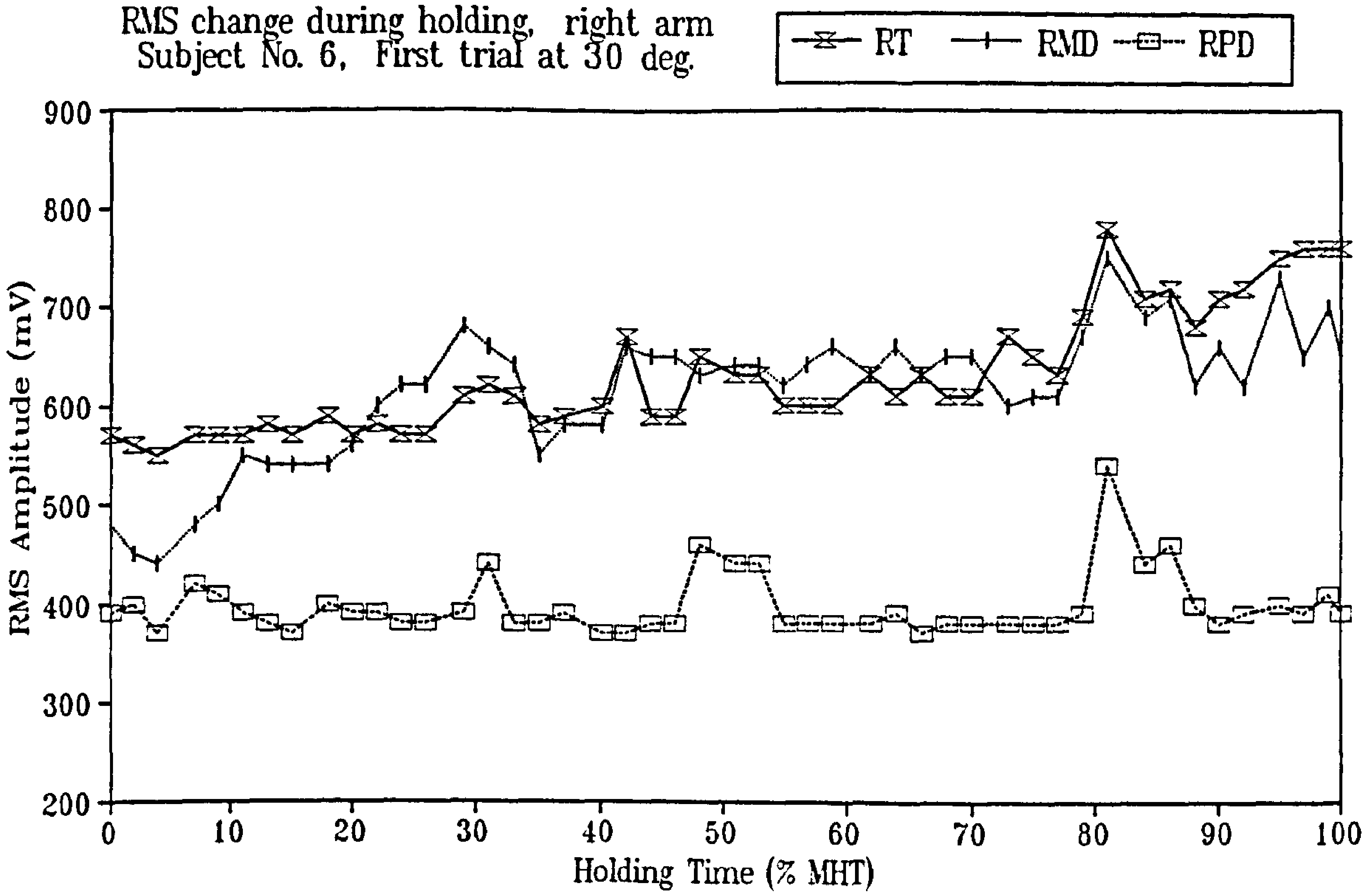


Fig D.5 Changes in RMS amplitude for subject No. 5 at 30 deg.



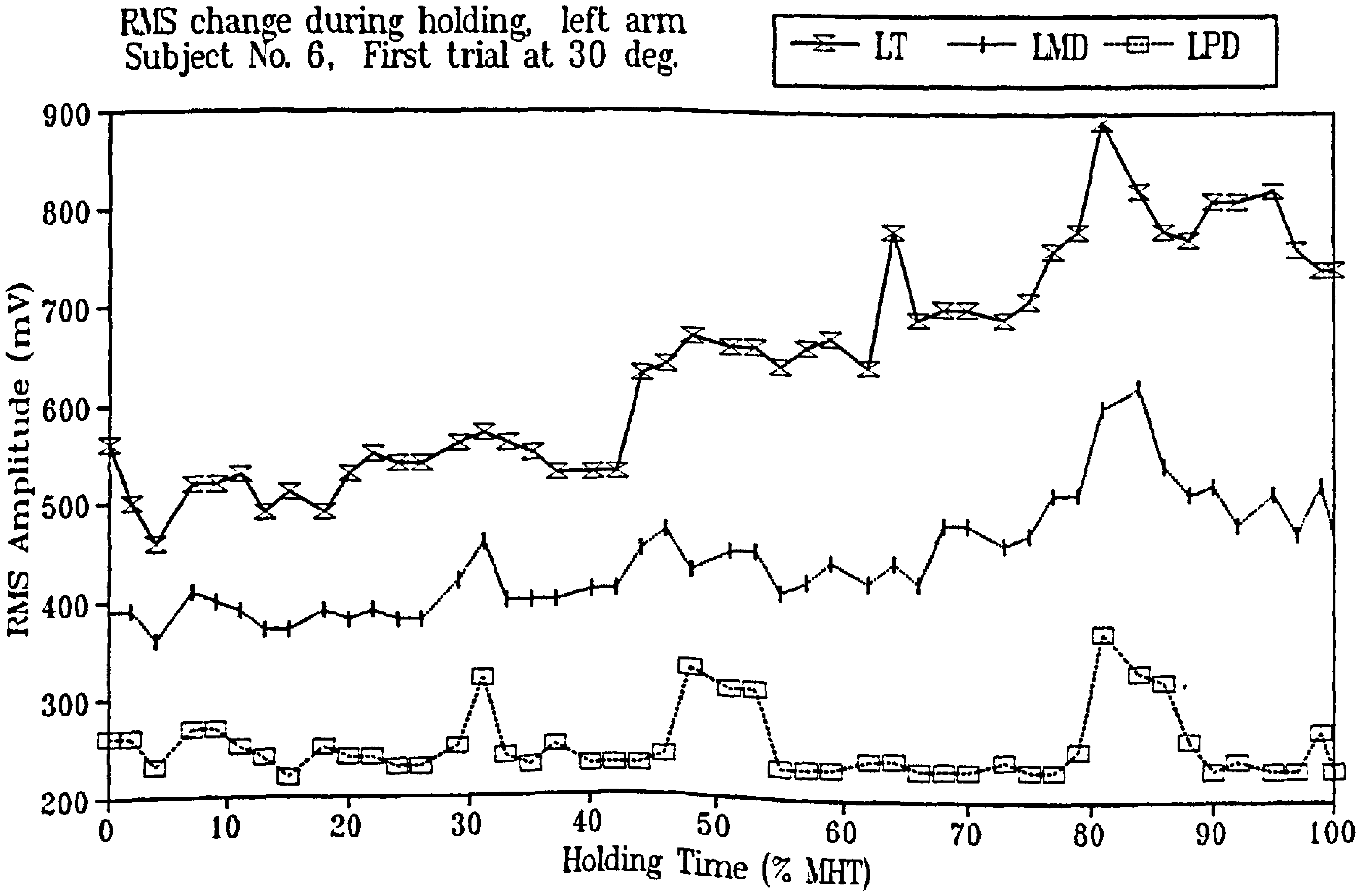
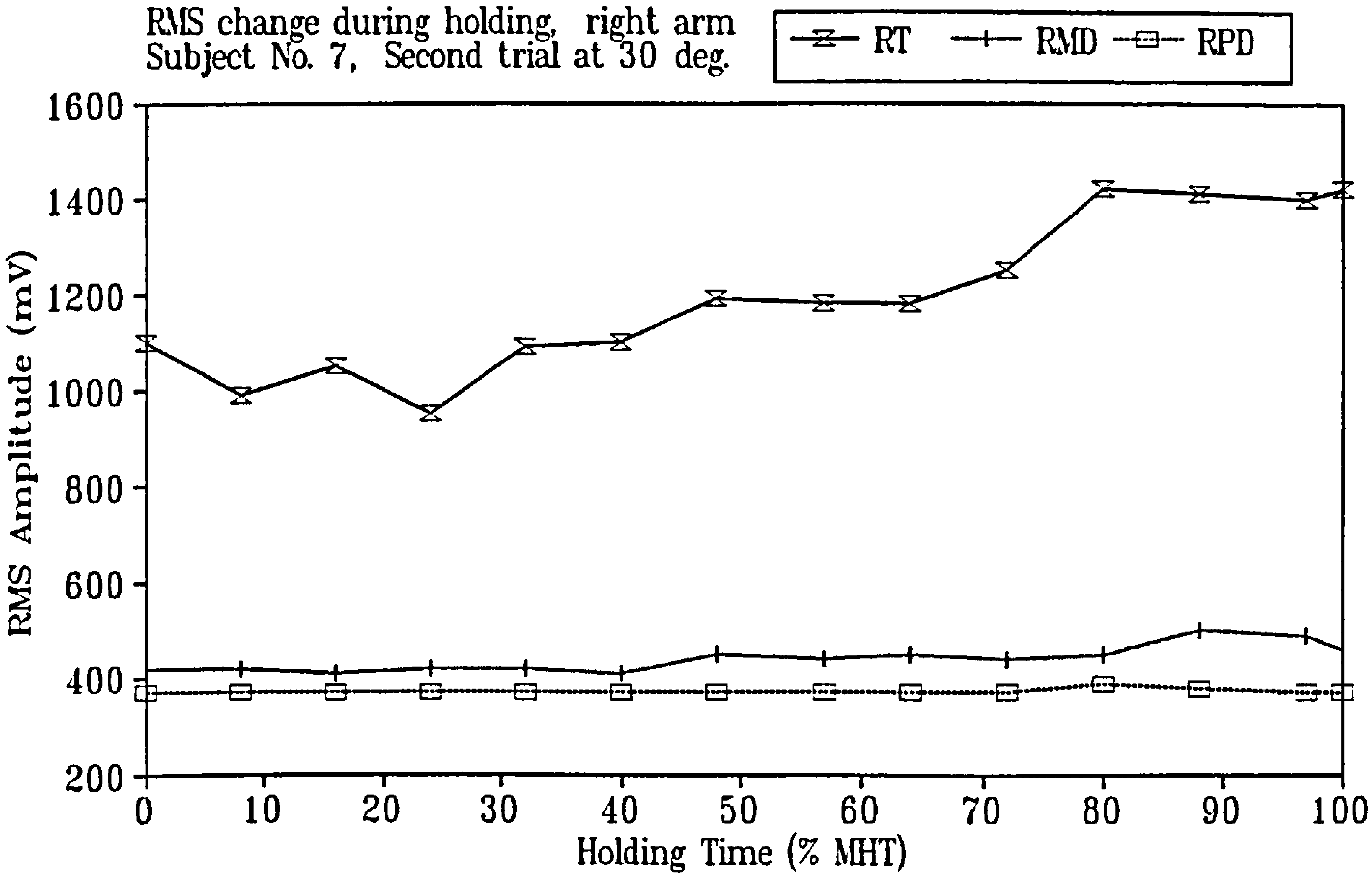


Figure D.6 Changes in RMS amplitude for subject 6 at 30 deg.



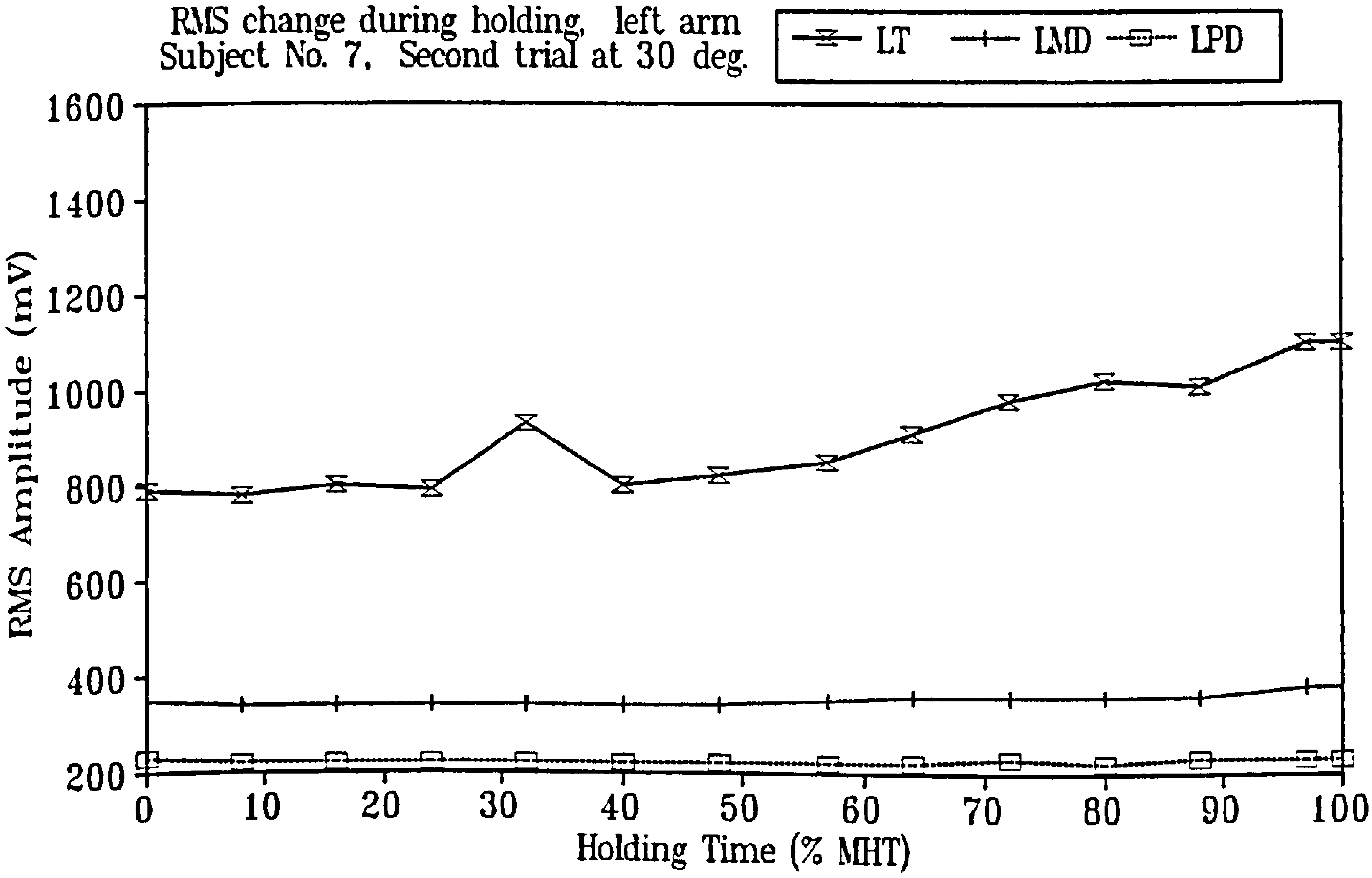
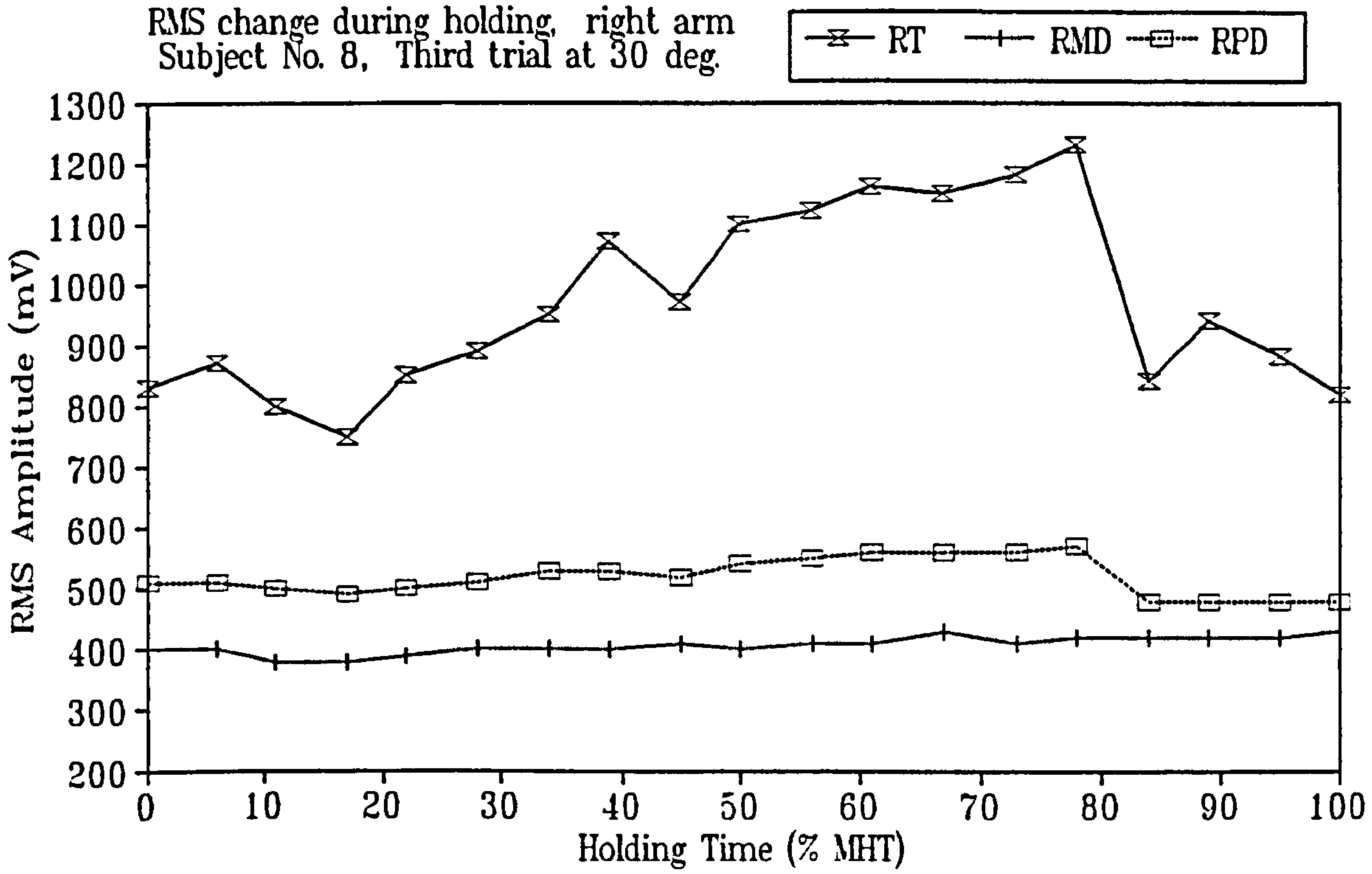


Fig D.7 Changes in RMS amplitude for subject No. 7 at 30 deg.



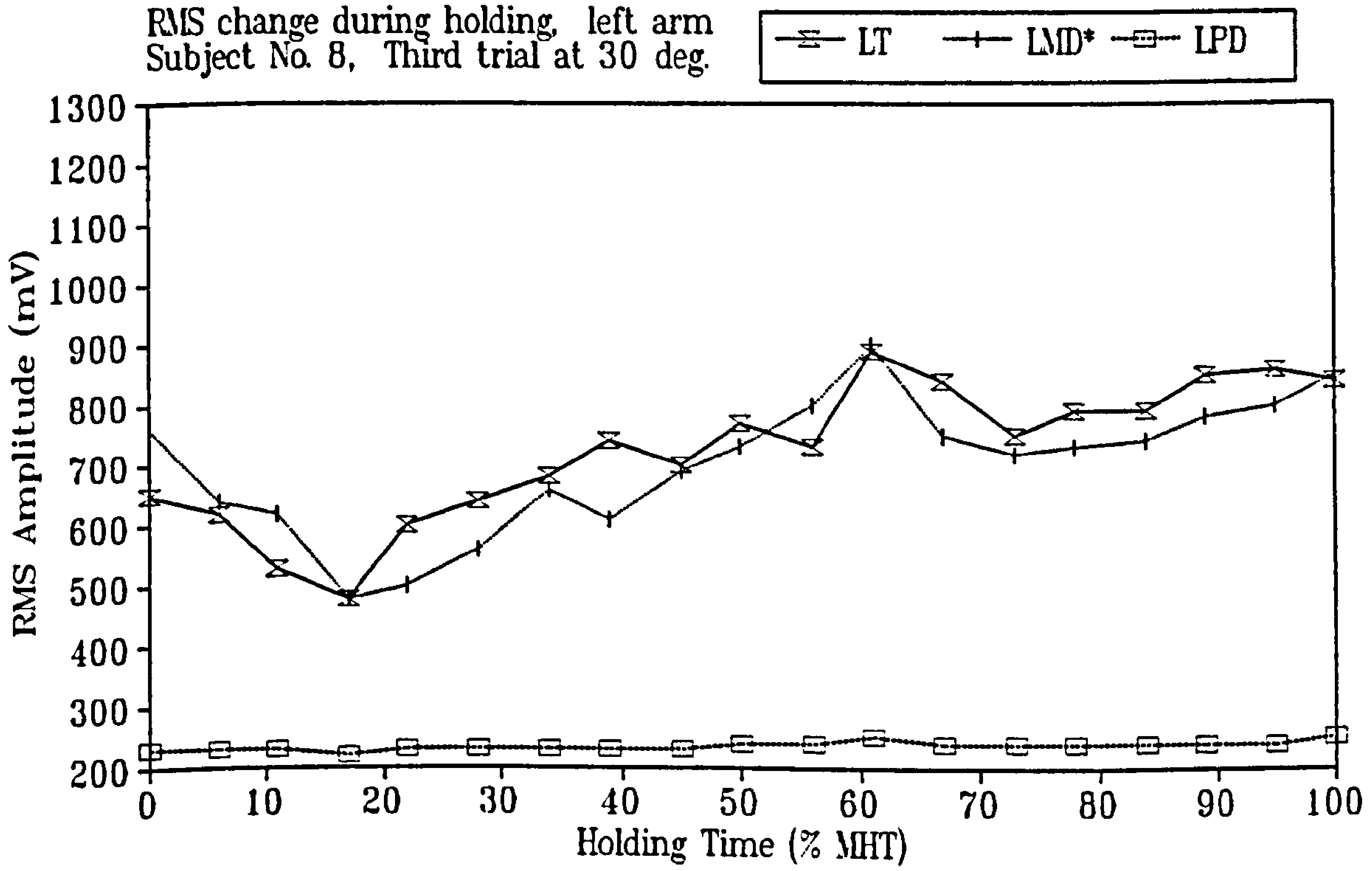


Fig D.8 Change in RMS amplitude for subject No. 8 at 30 deg.

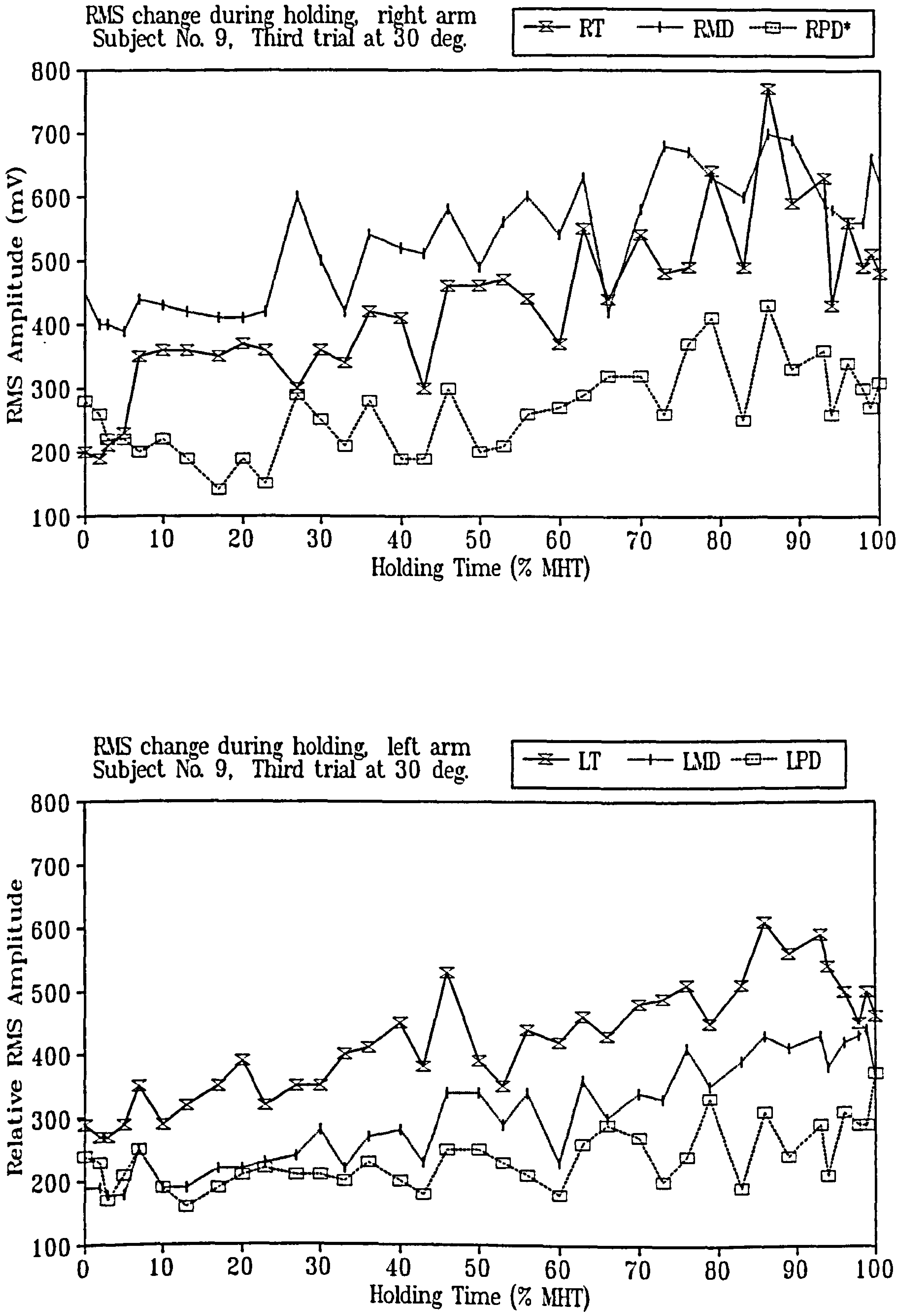
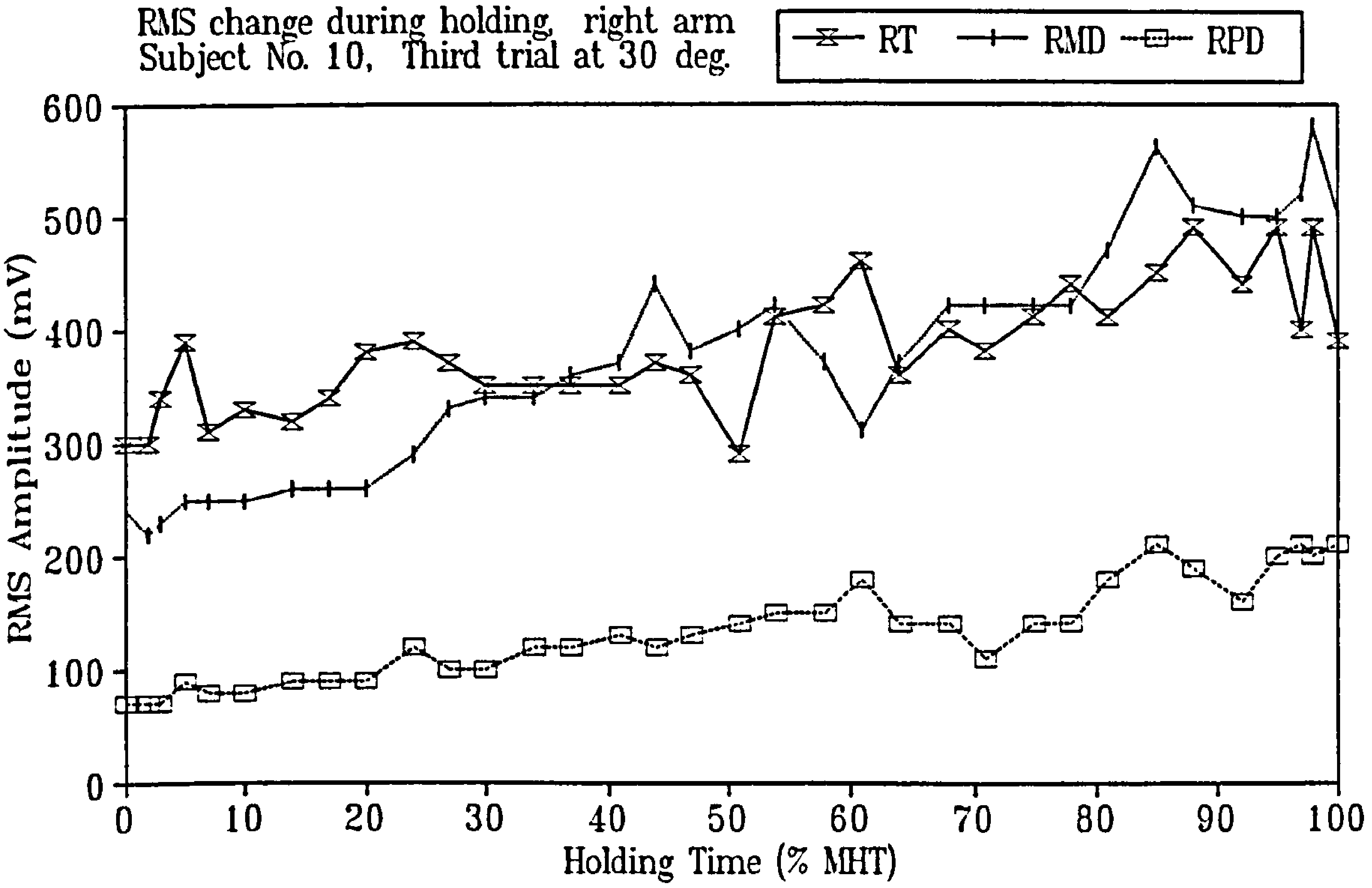


Figure D.9 Changes in RMS amplitude for subject No. 9 at 30 deg.



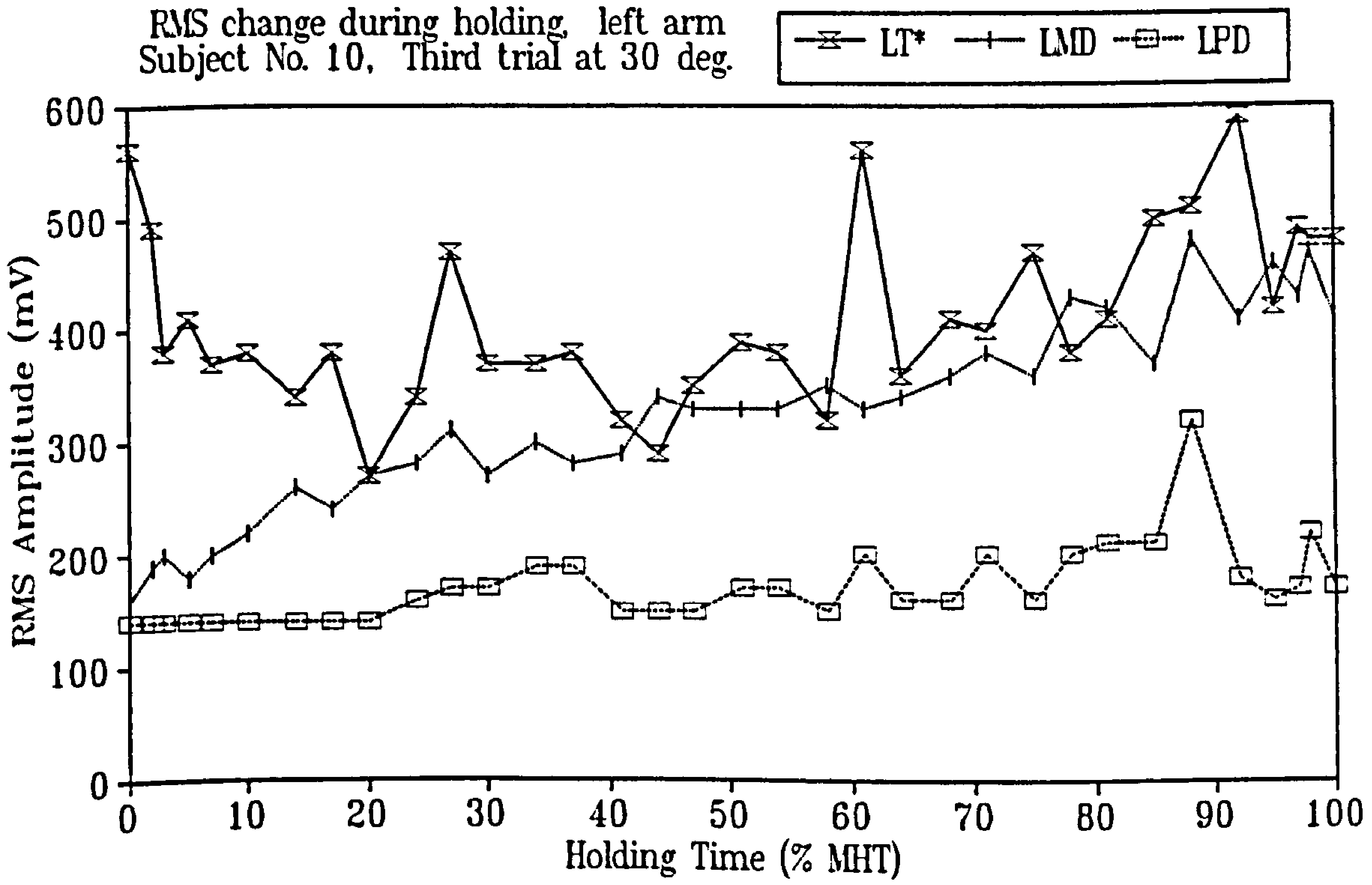
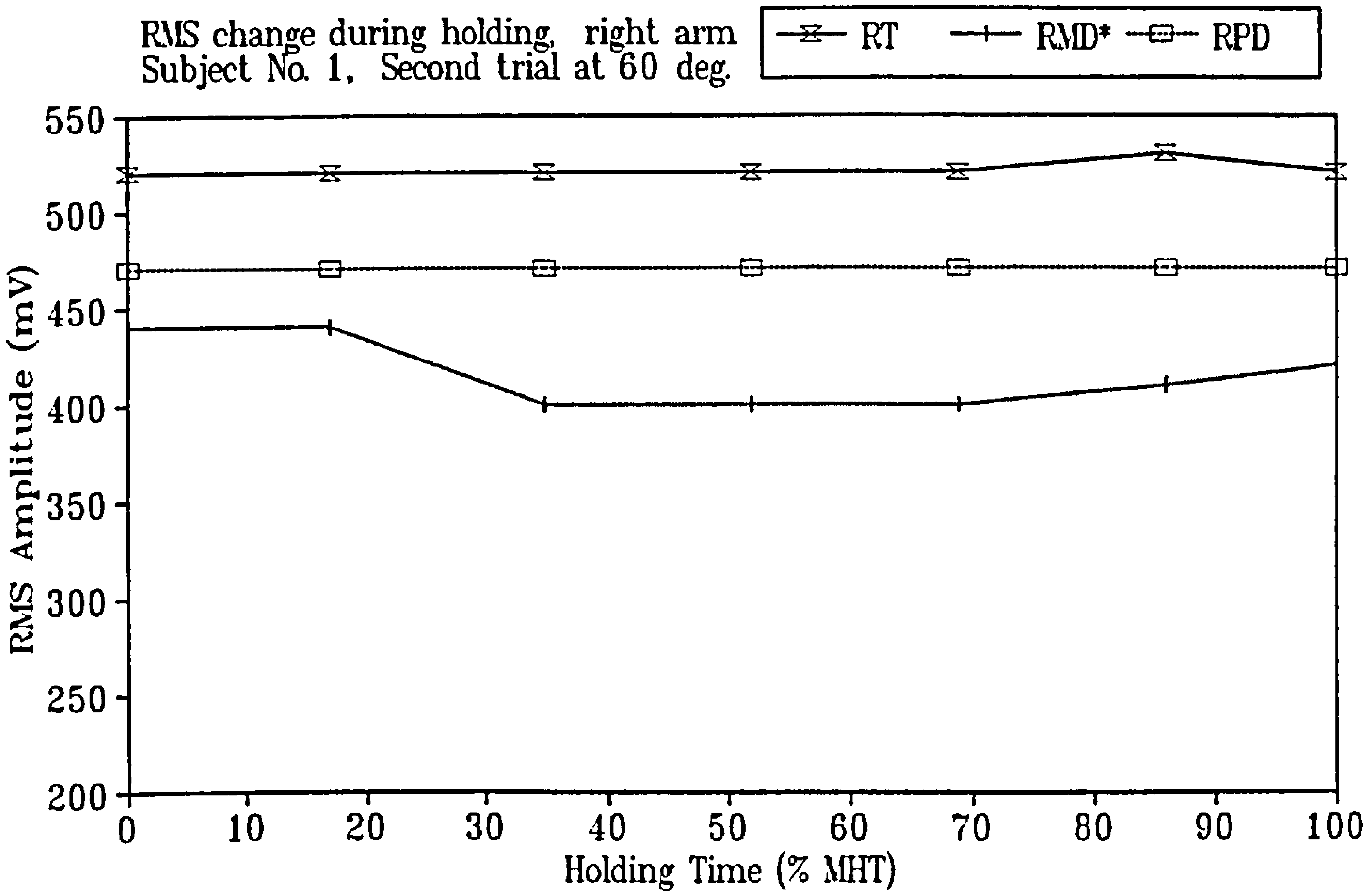


Figure D.10 Changes in RMS amplitude for subject No. 10 at 30 deg.



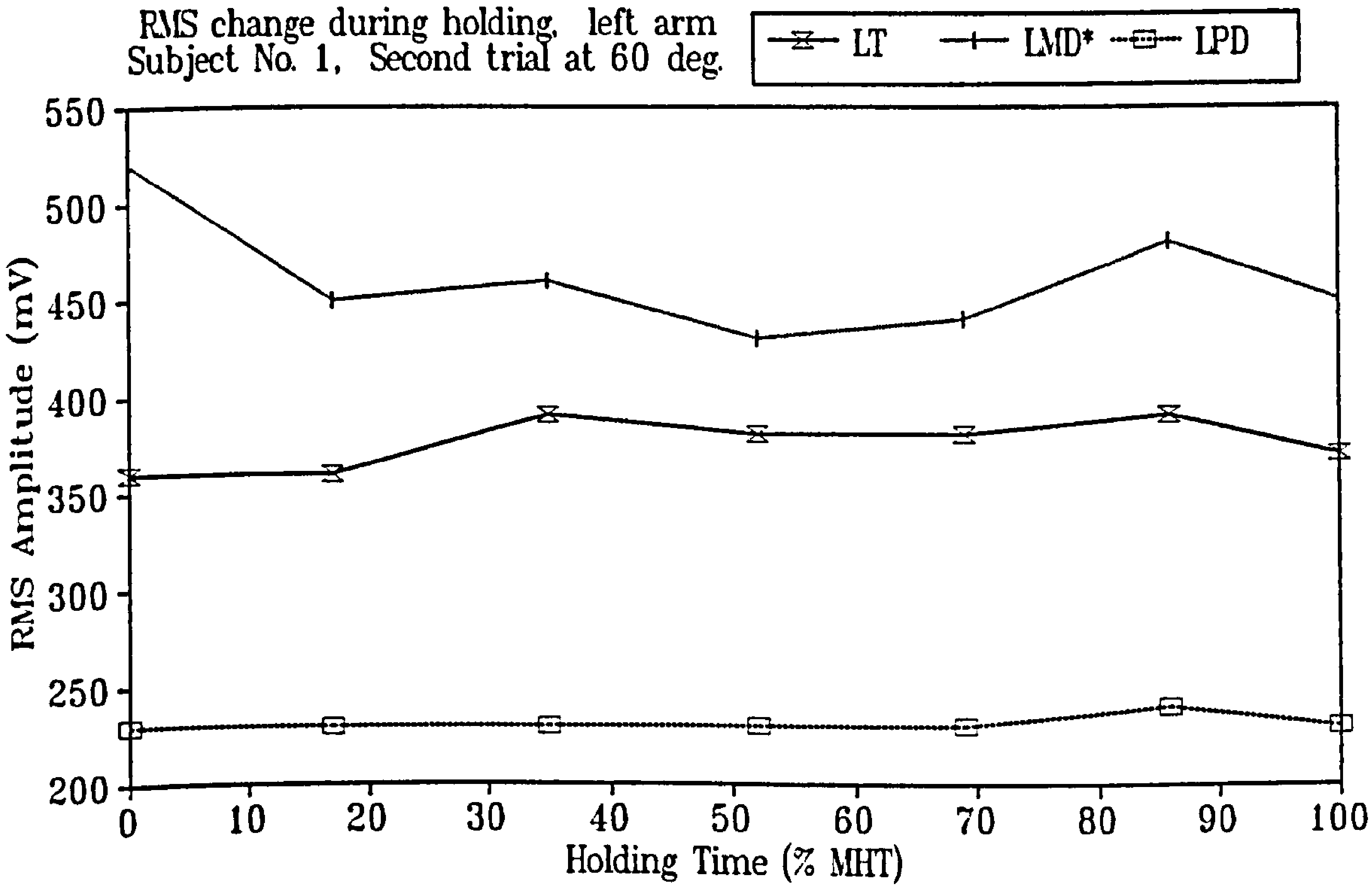
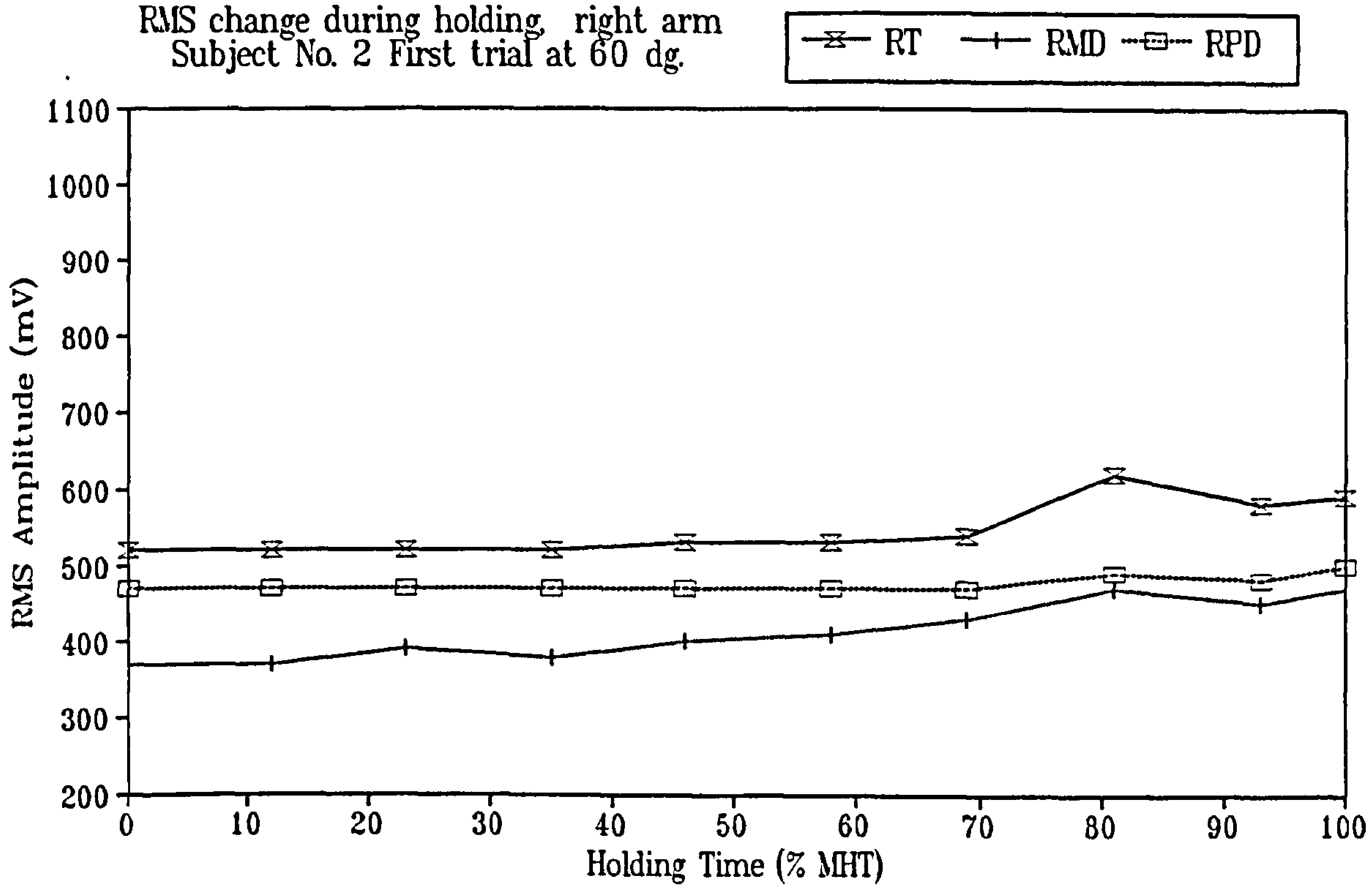


Figure D.11 Changes in RMS amplitude for subject No. 1 at 60 deg.



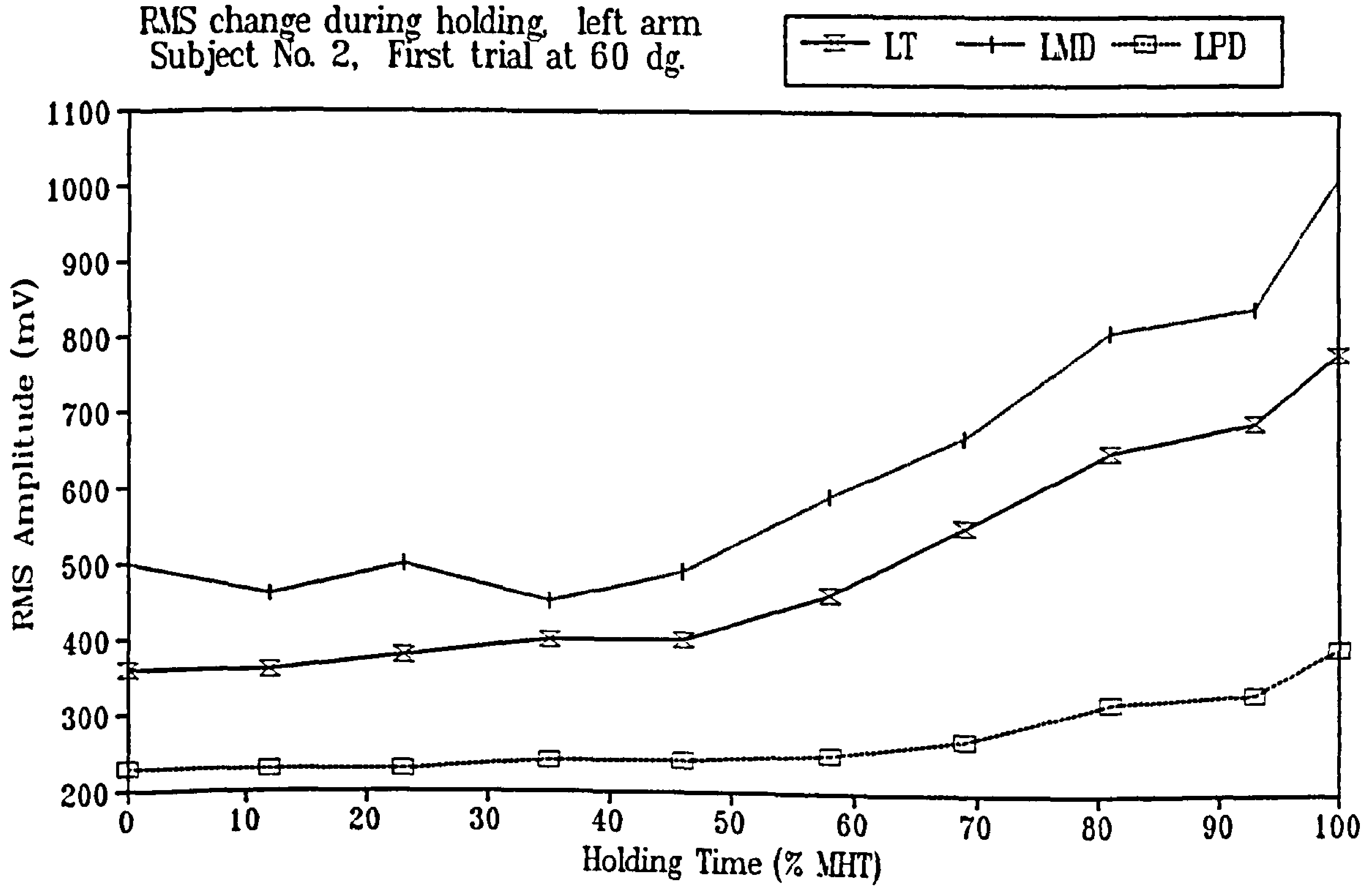
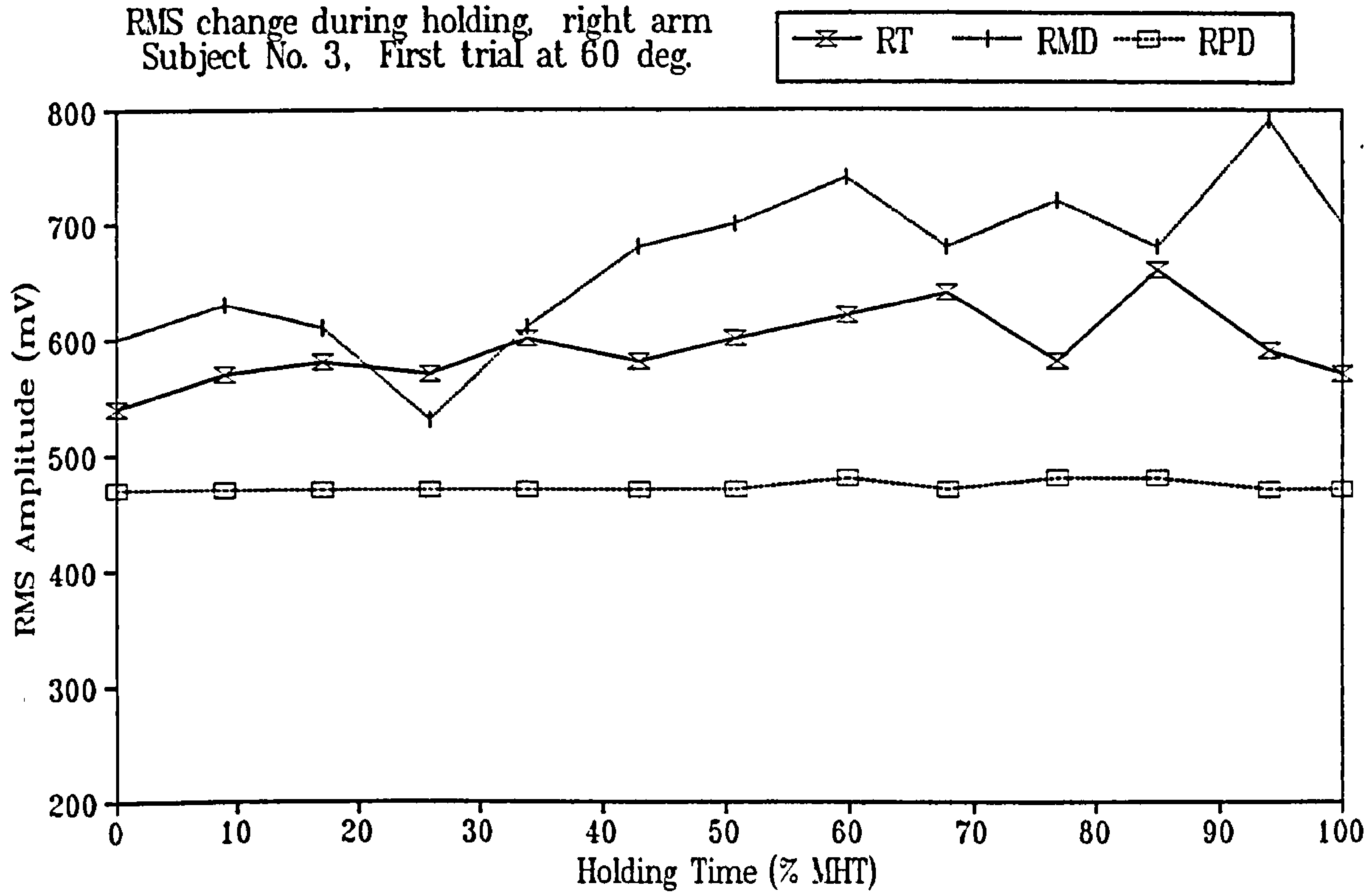


Figure D.12 Changes in RMS amplitude for subject No. 2 at 60 dg.



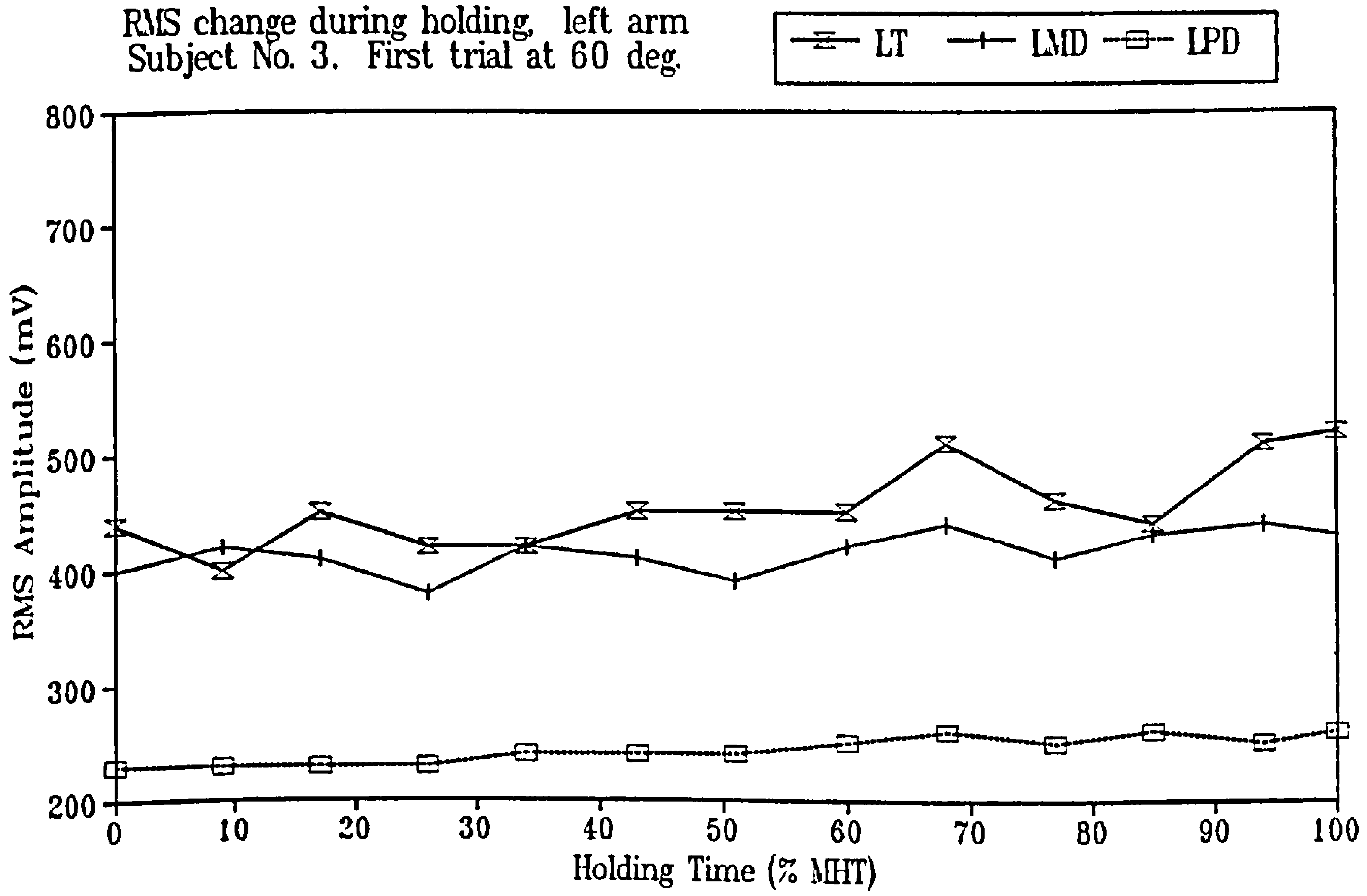
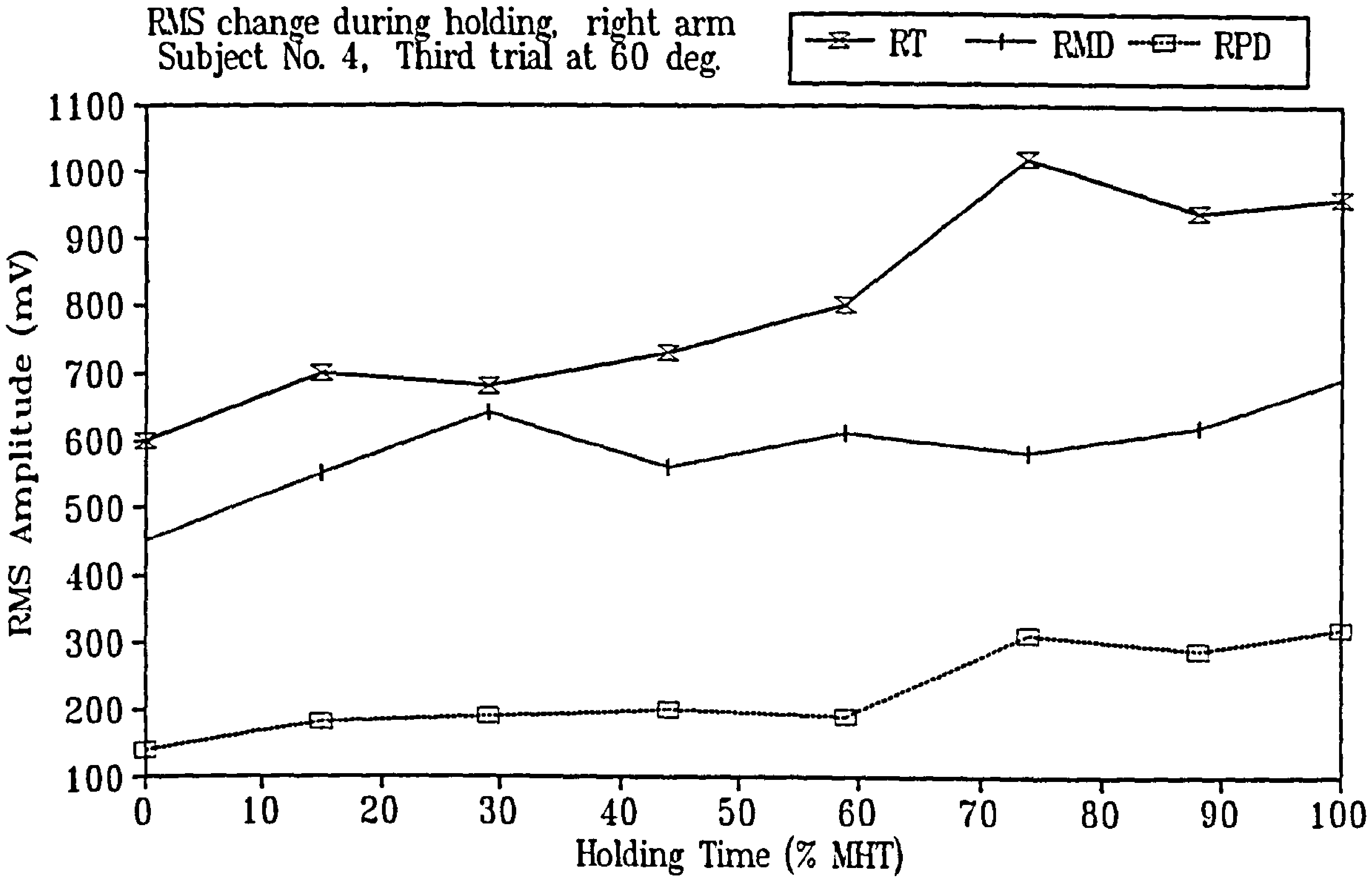


Figure D.13 Changes in RMS amplitude for subject No. 3 at 60 dg.



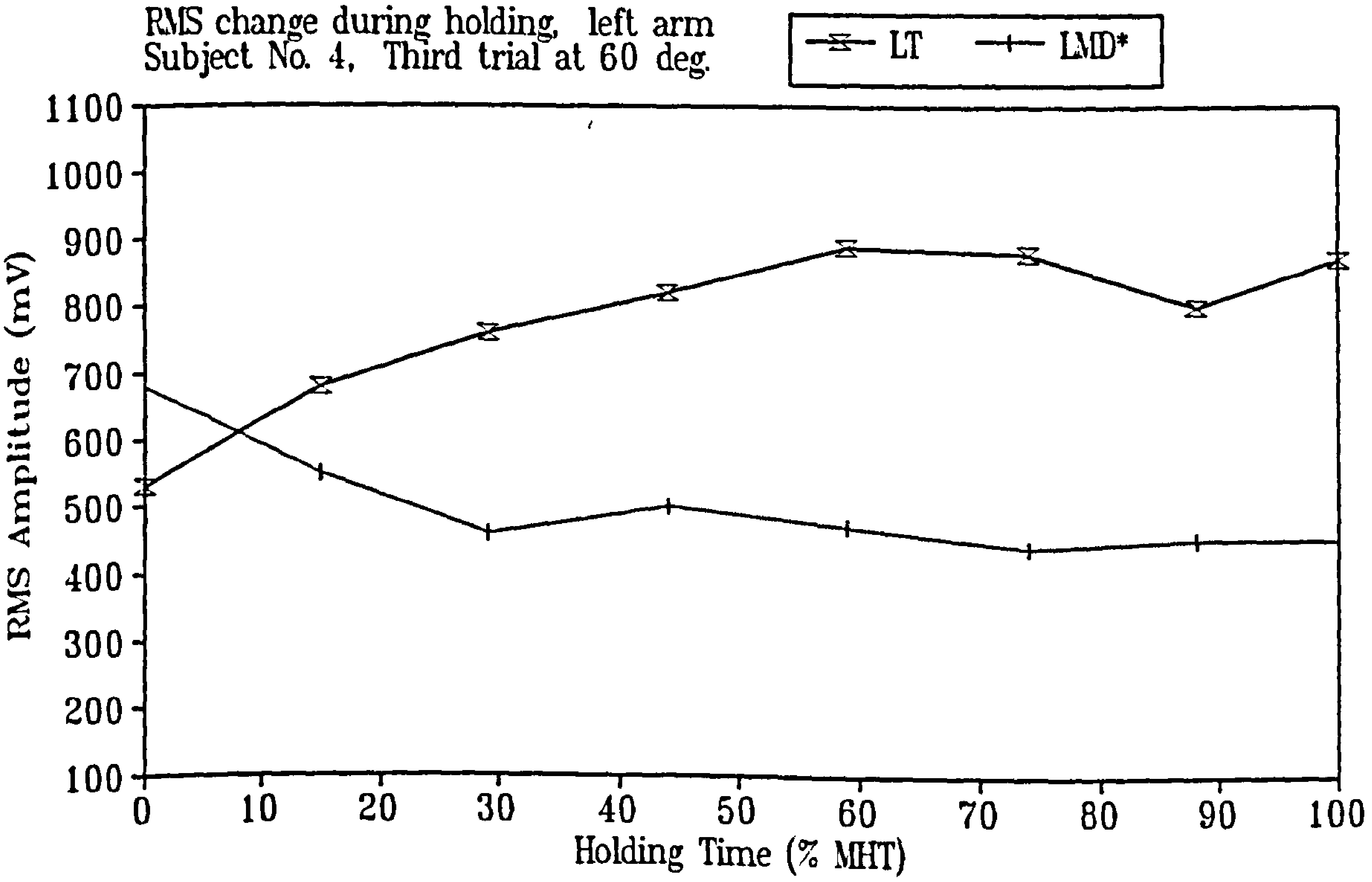
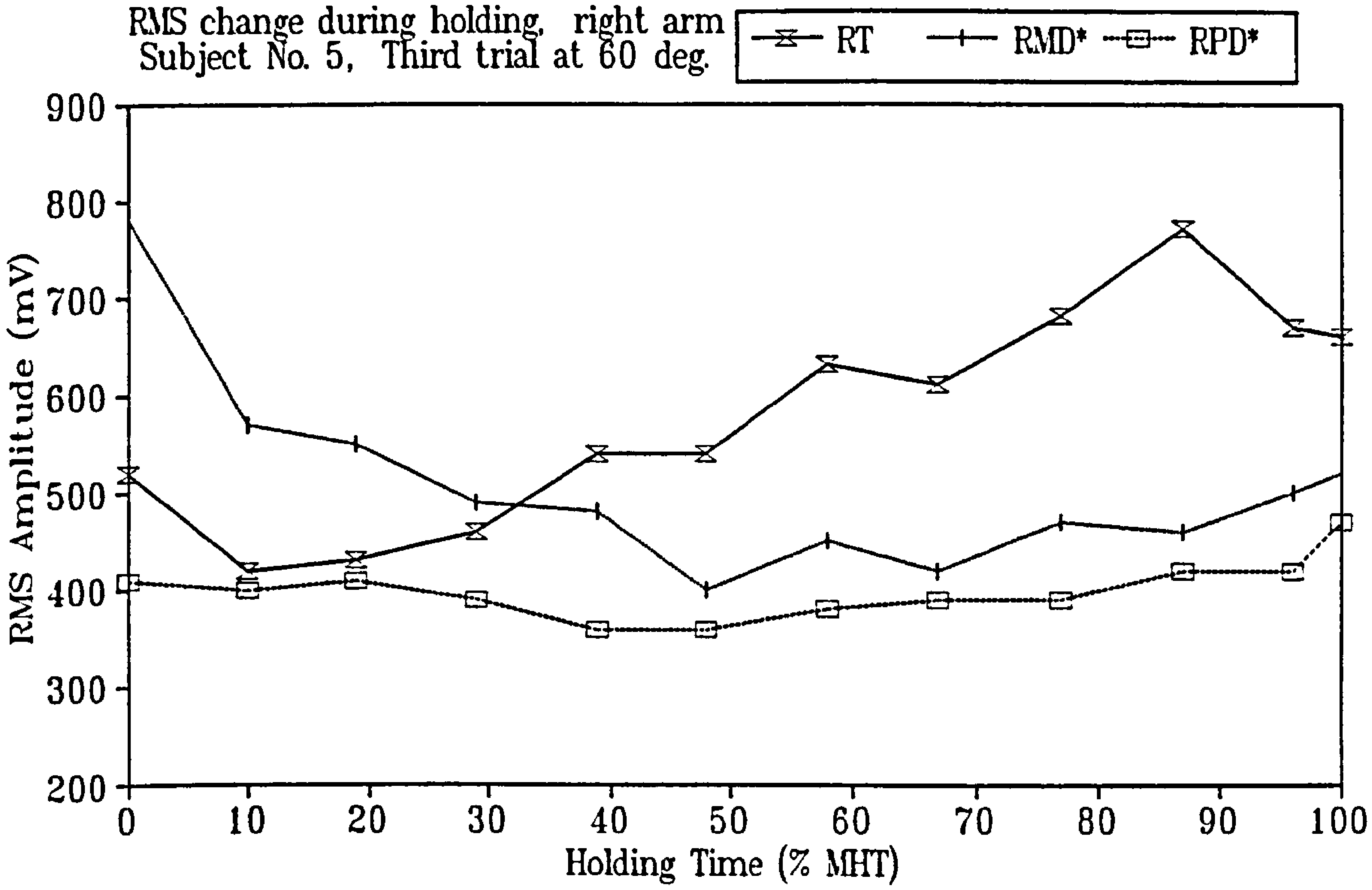


Fig D.14 Changes in RMS amplitude for subject No. 4 at 60 deg.



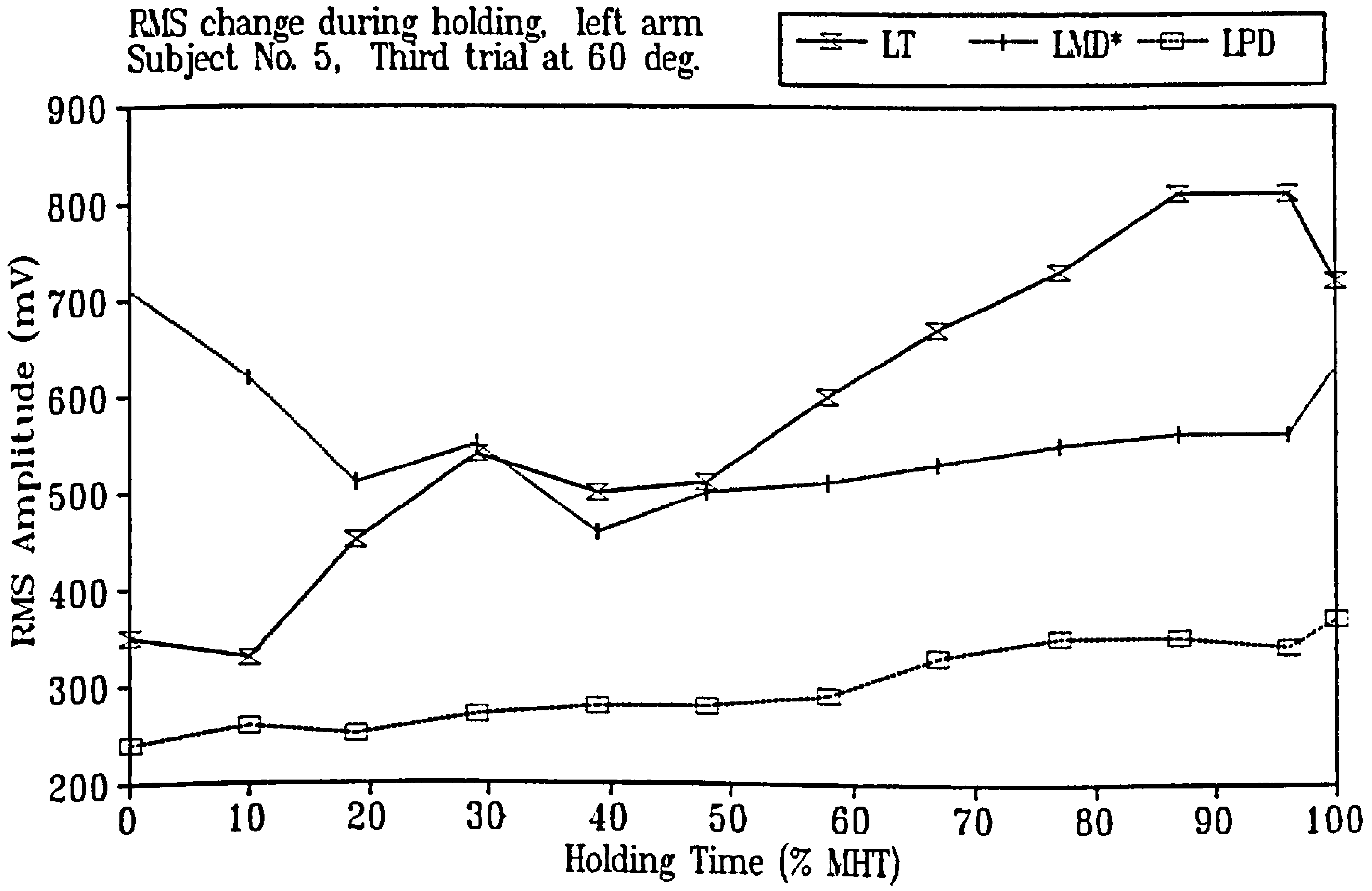
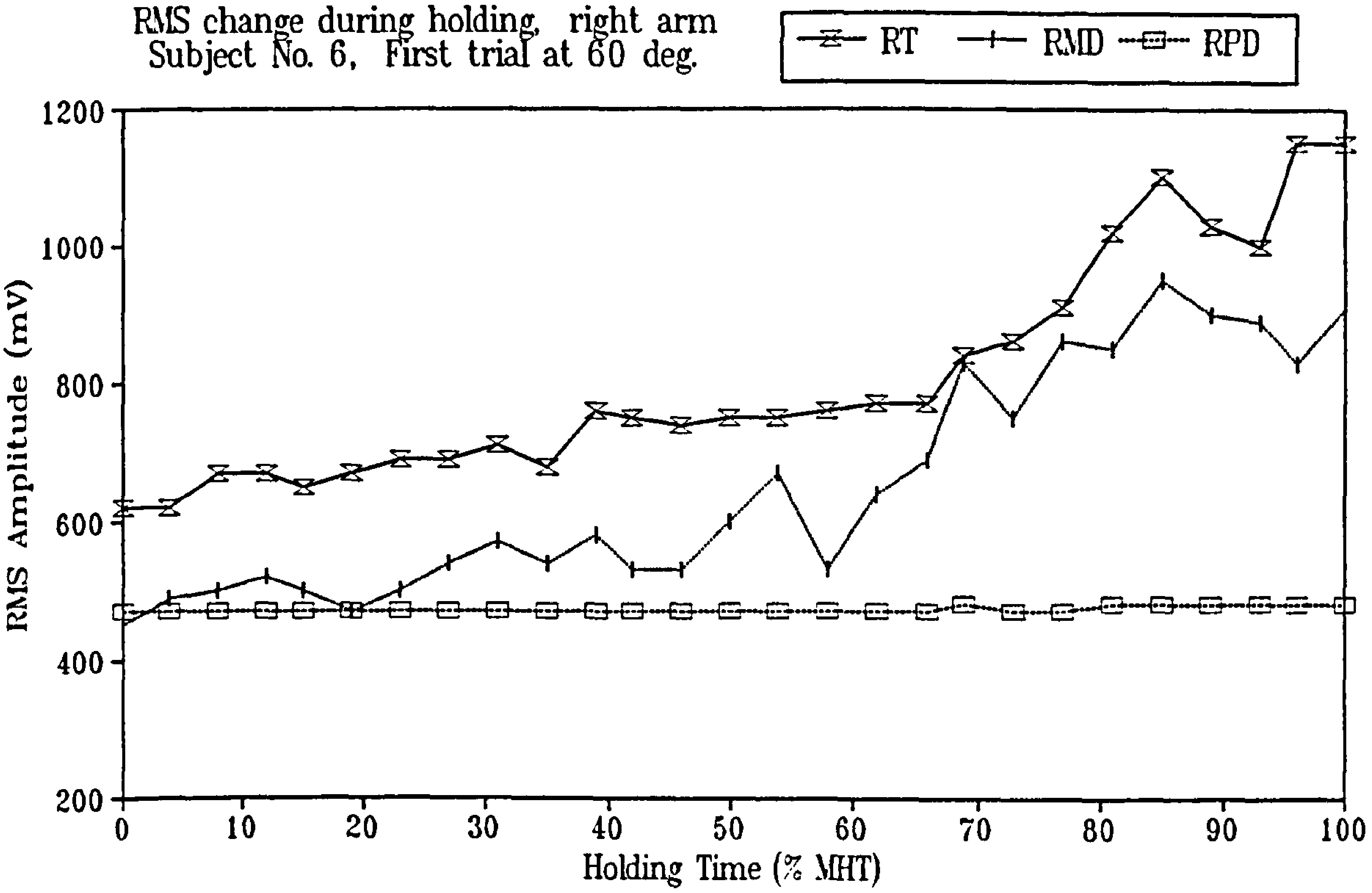


Figure D.15 Changes in RMS amplitude for subject No. 5 at 60 deg.



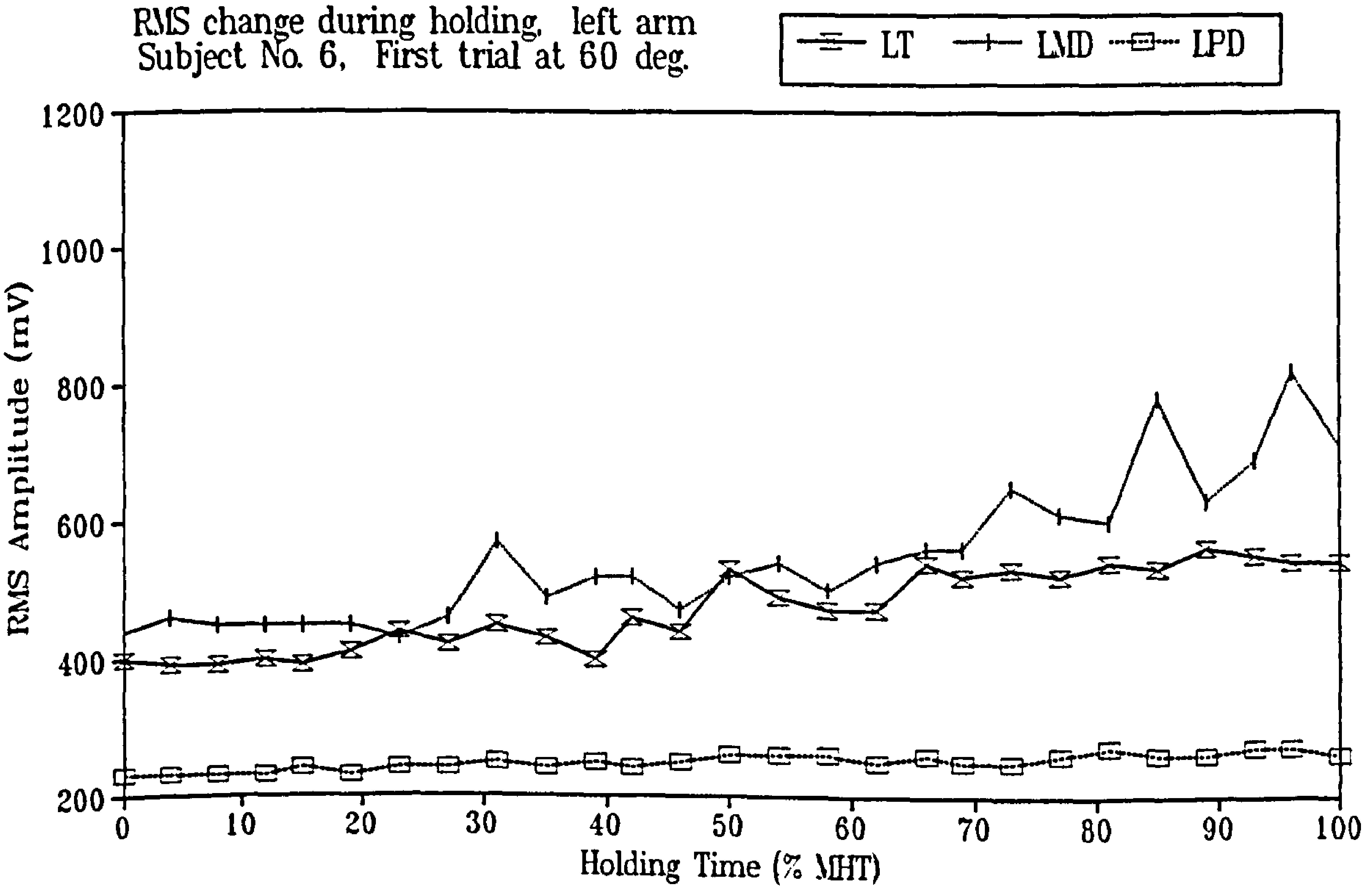
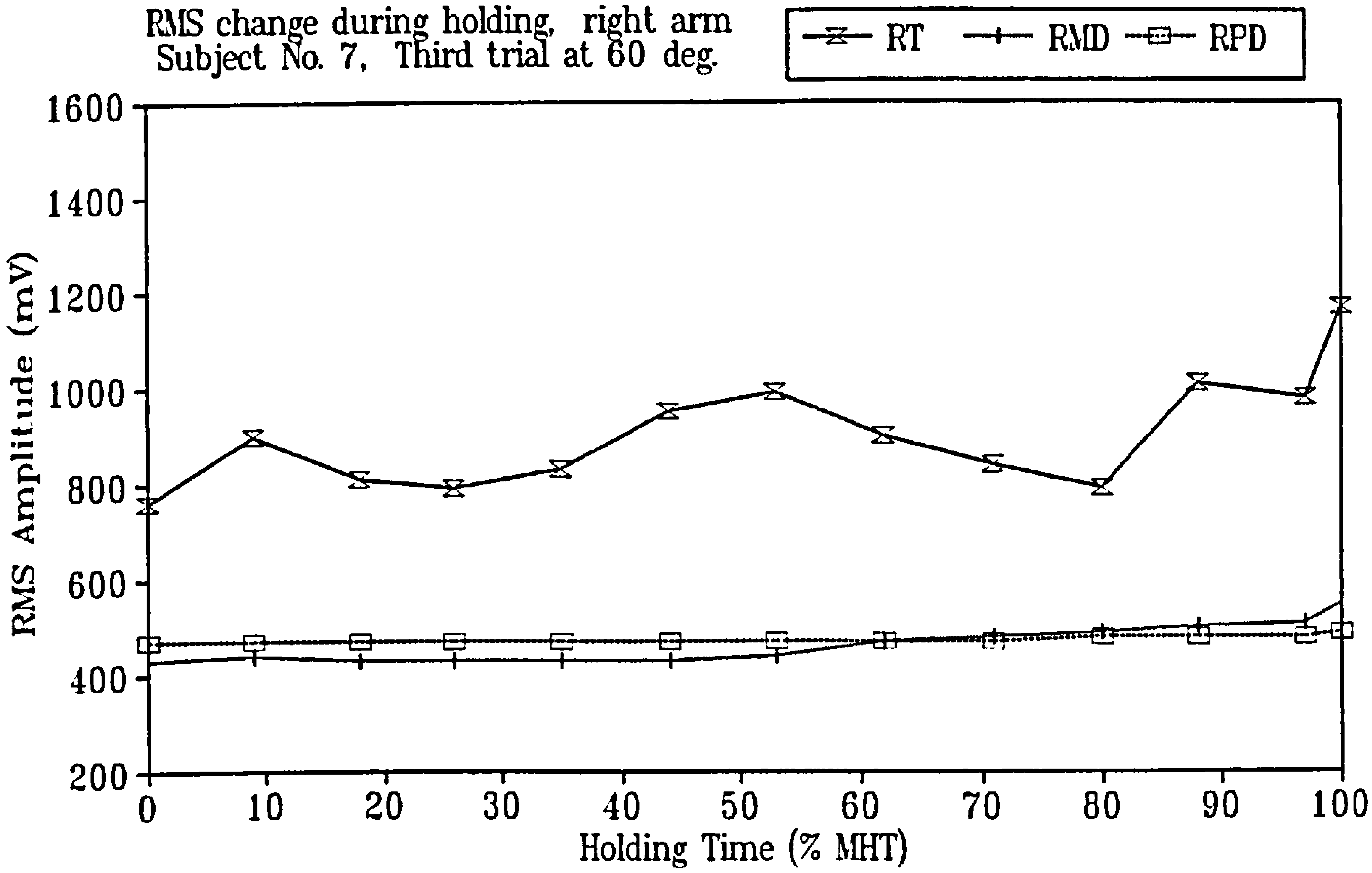


Figure D.16 Changes in RMS amplitude for subject No. 6 at 60 deg.



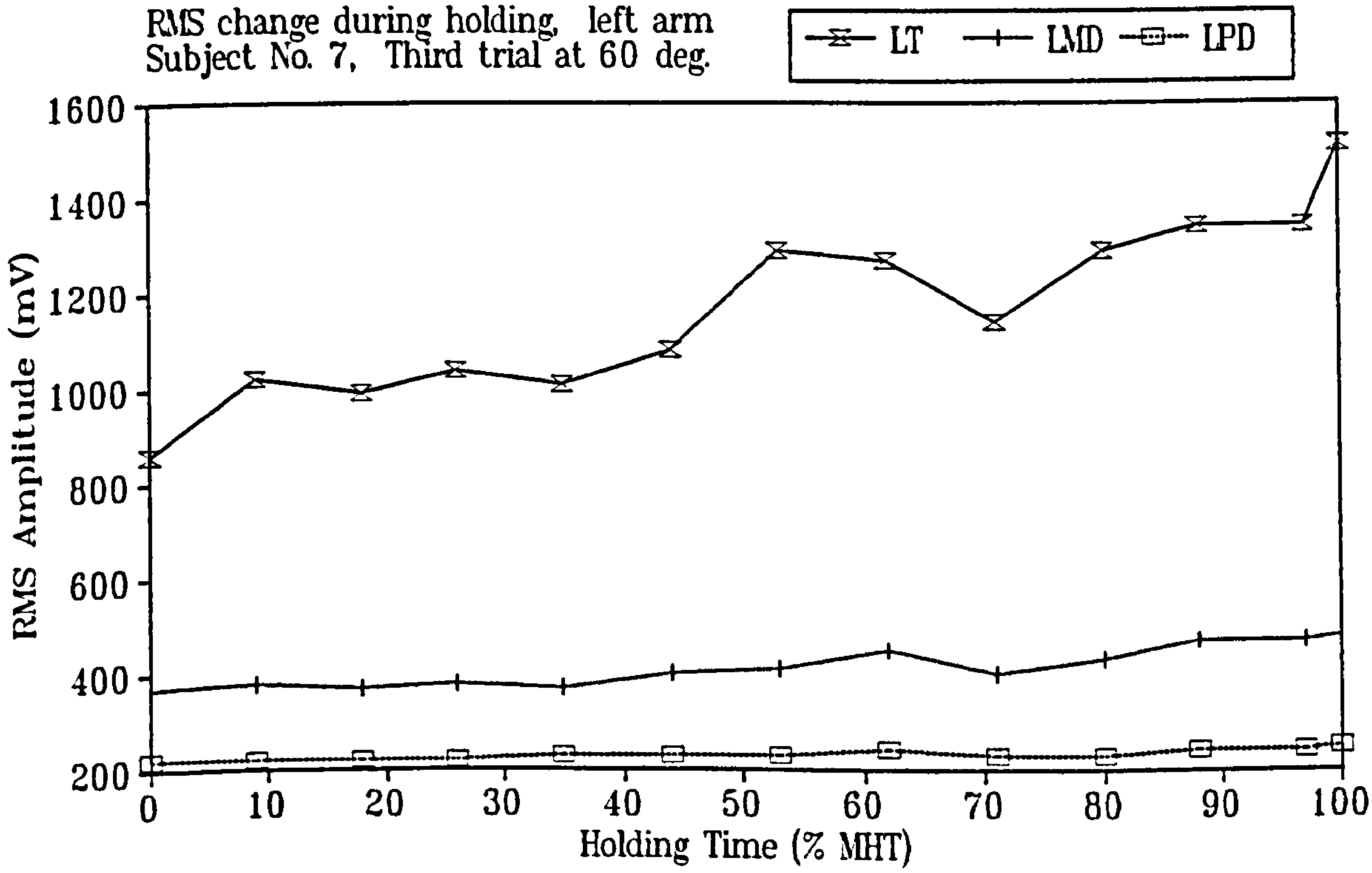


Figure D.17 Changes in RMS amplitude for subject No. 7 at 60 deg.

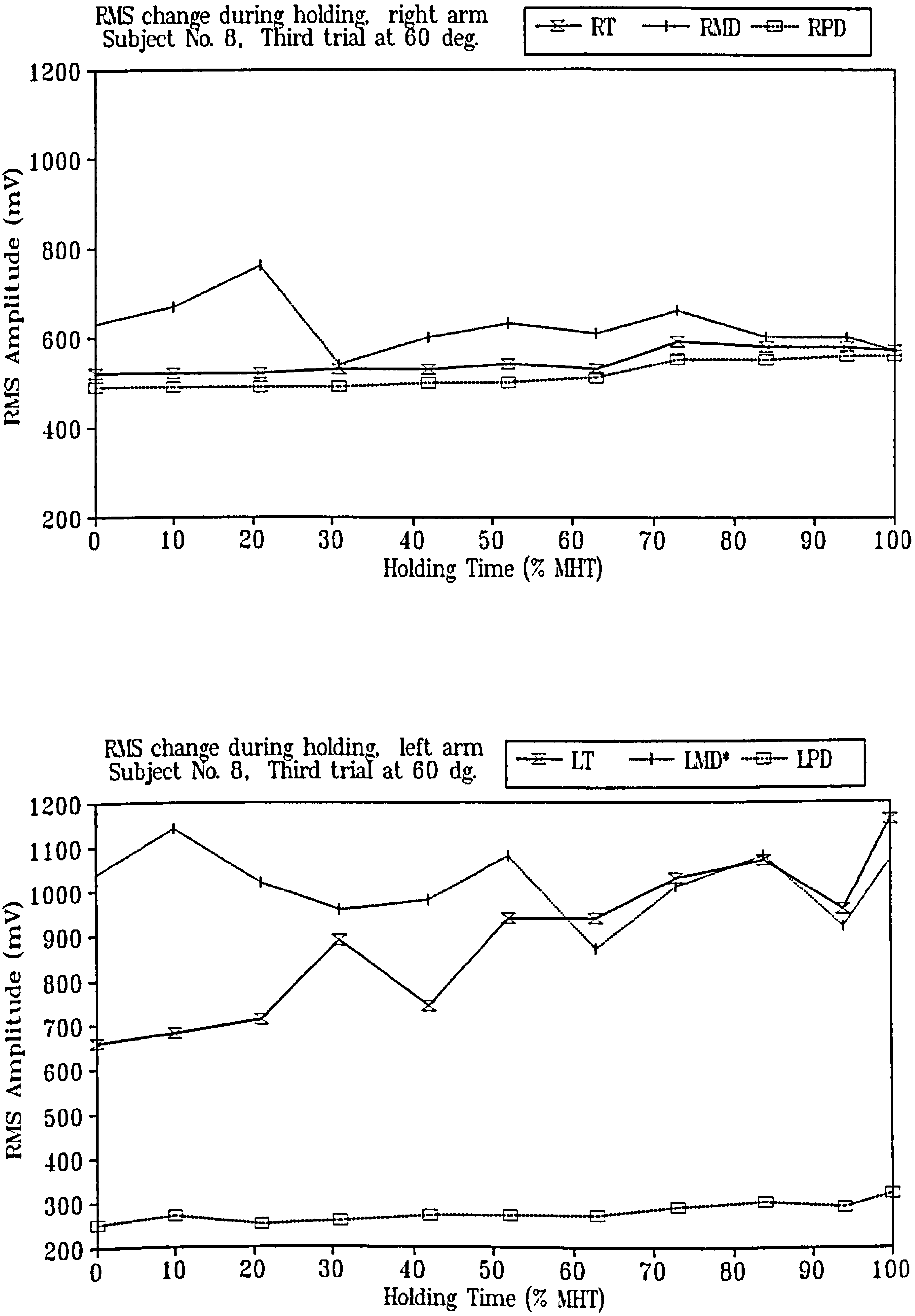
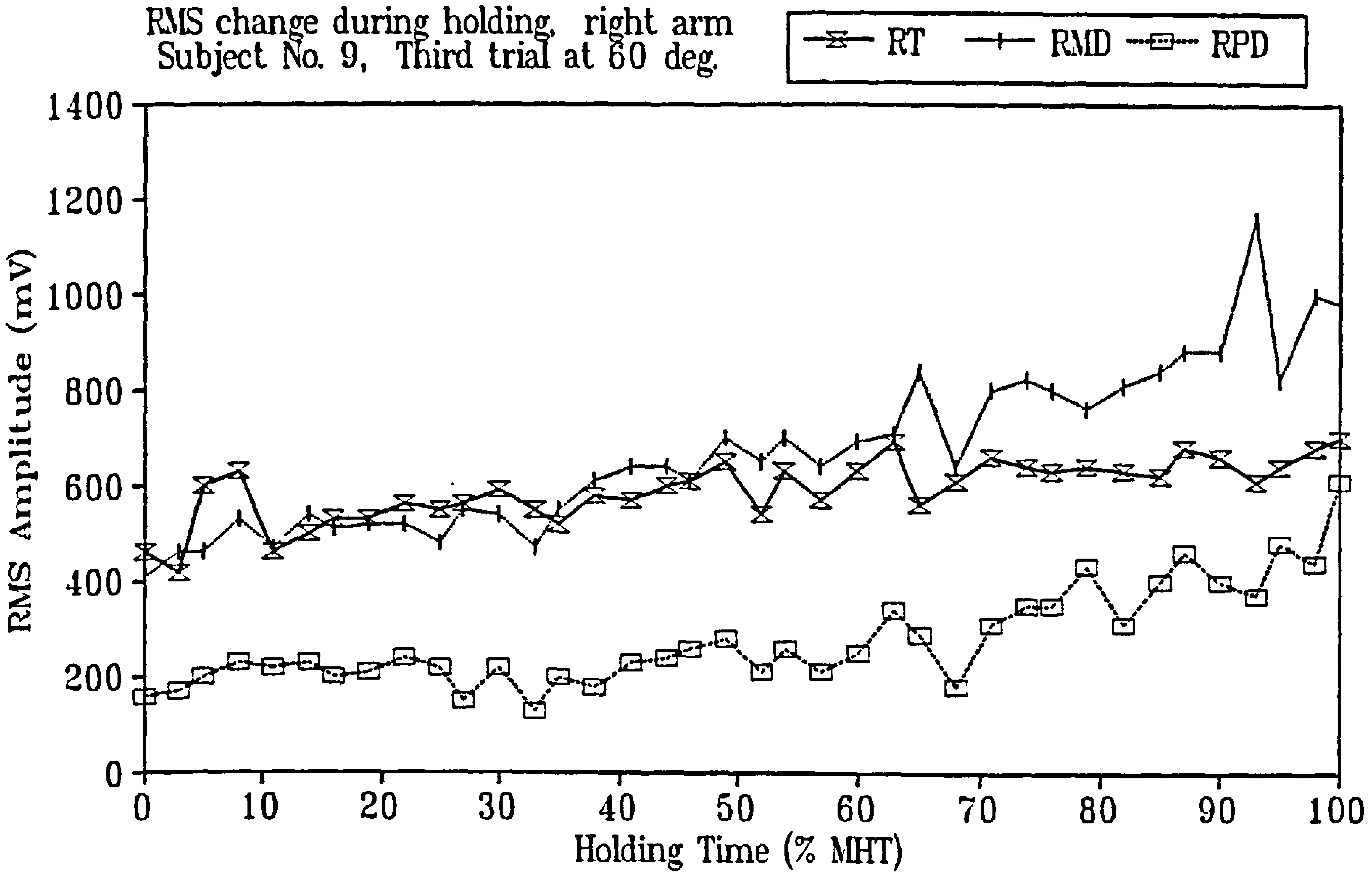


Figure D.18 Changes in RMS amplitude for subject No. 8 at 60 deg.



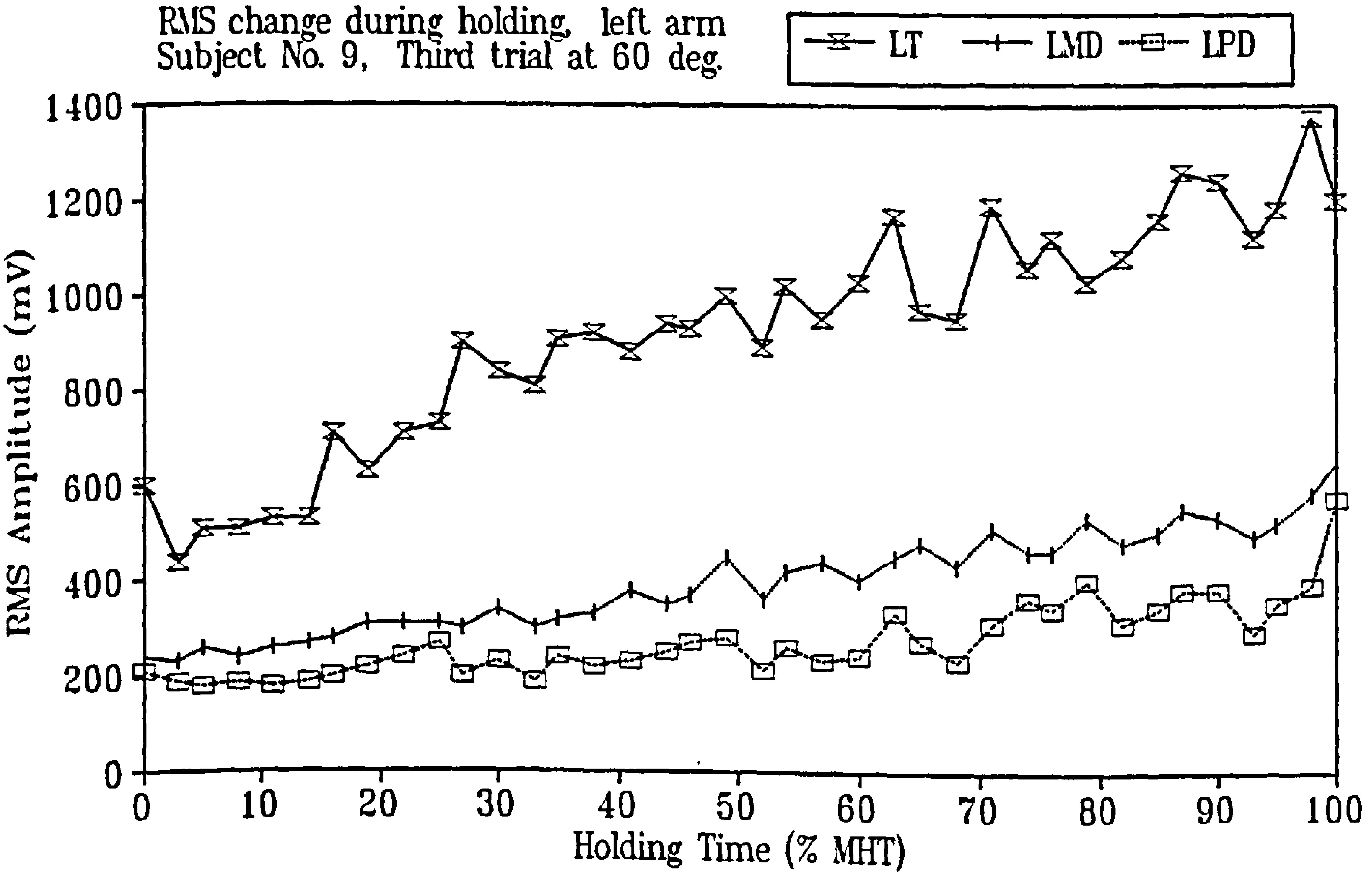
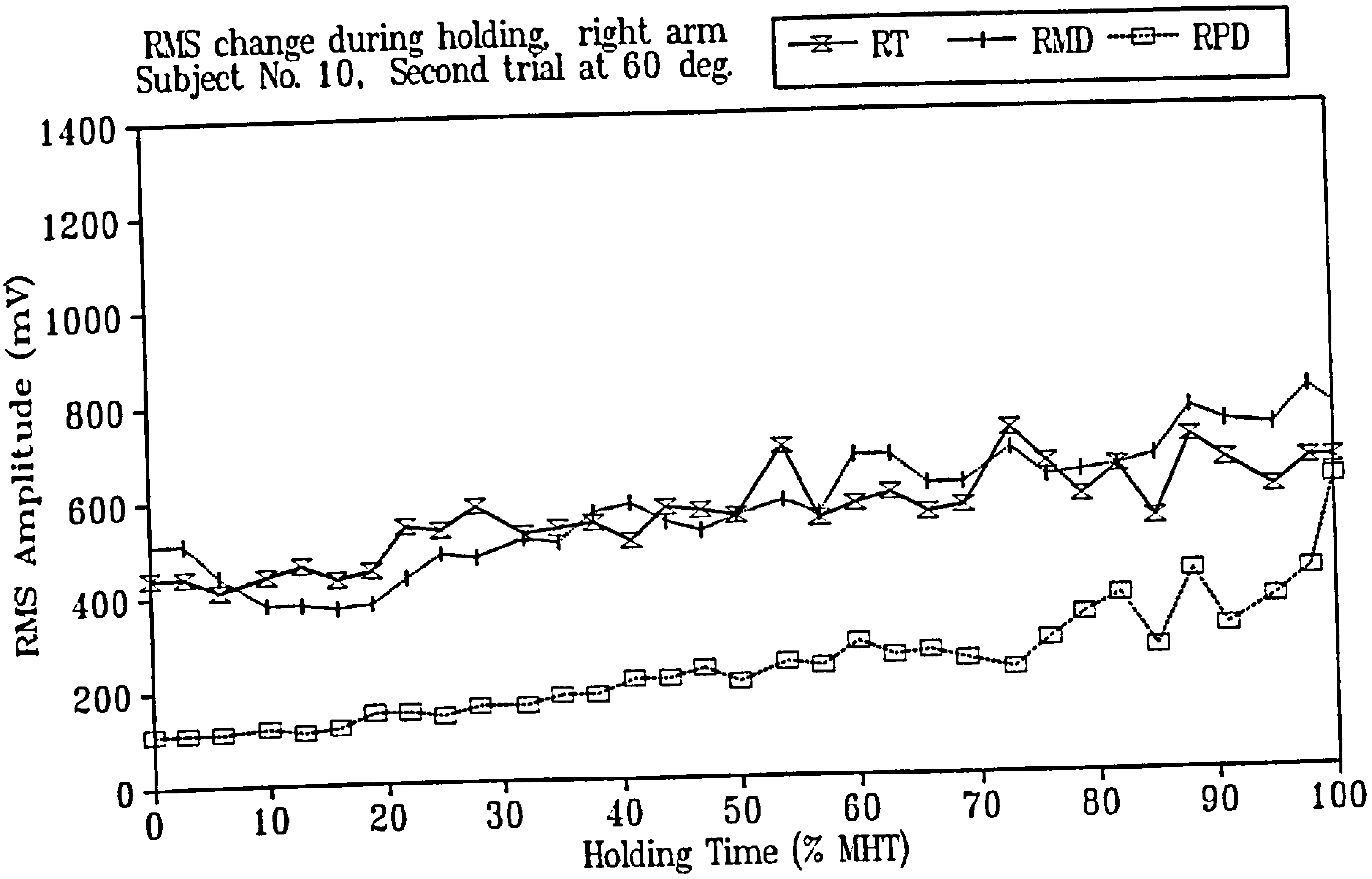


Figure D.19 Changes in RMS amplitude for subject No. 9 at 60 deg.



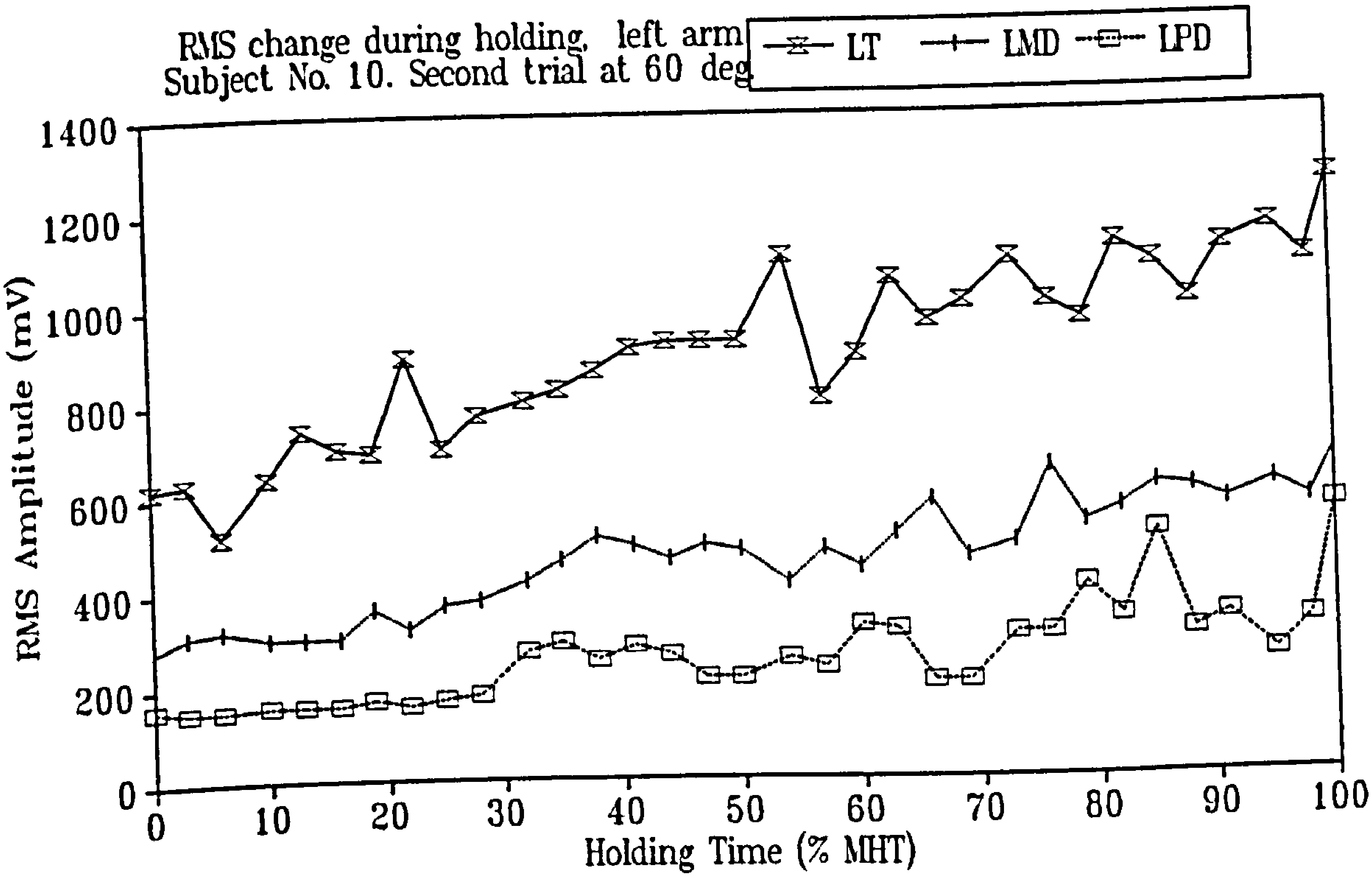
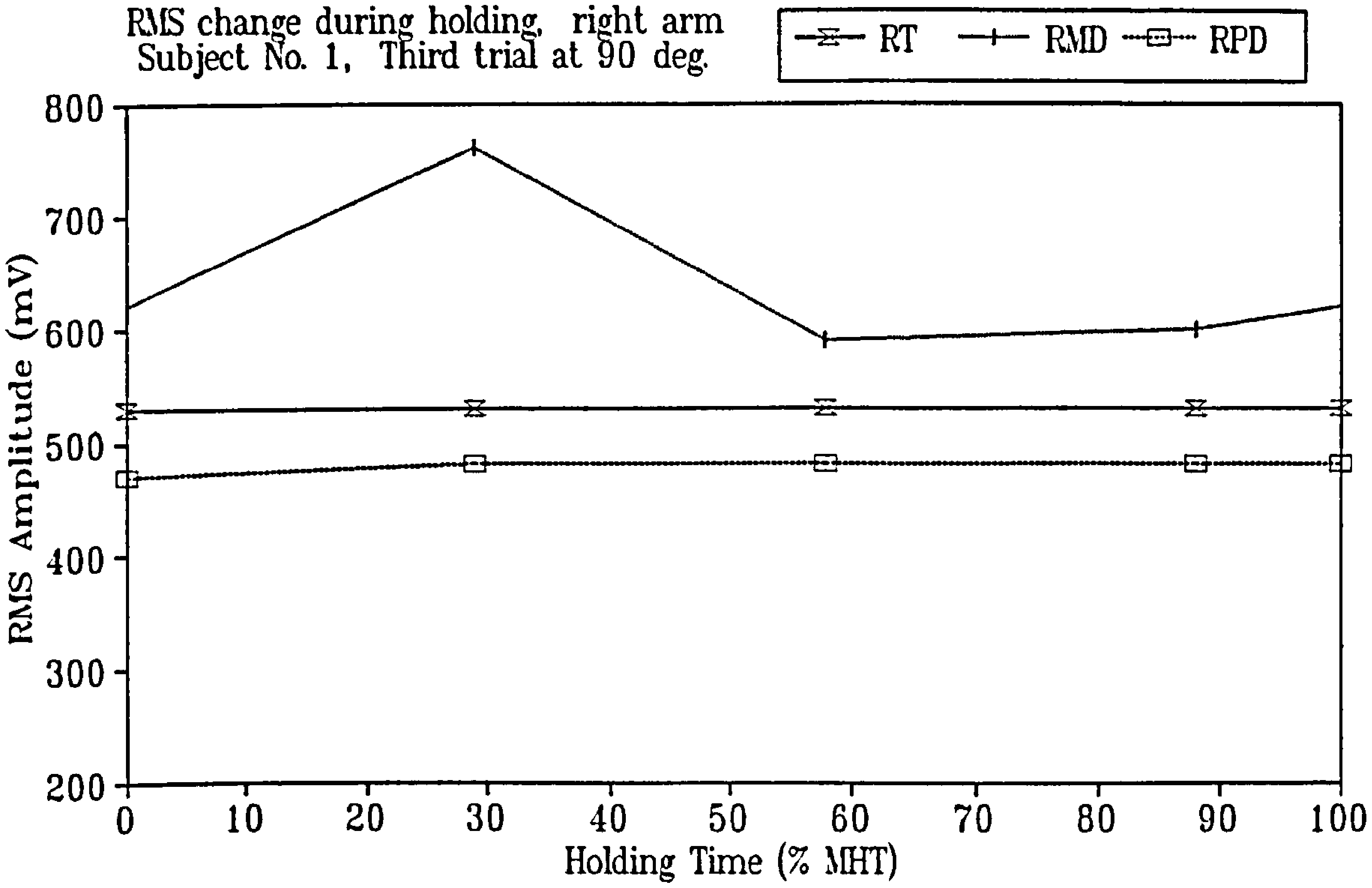


Figure D.20 Changes in RMS amplitude for subject 10 at 60 dg.



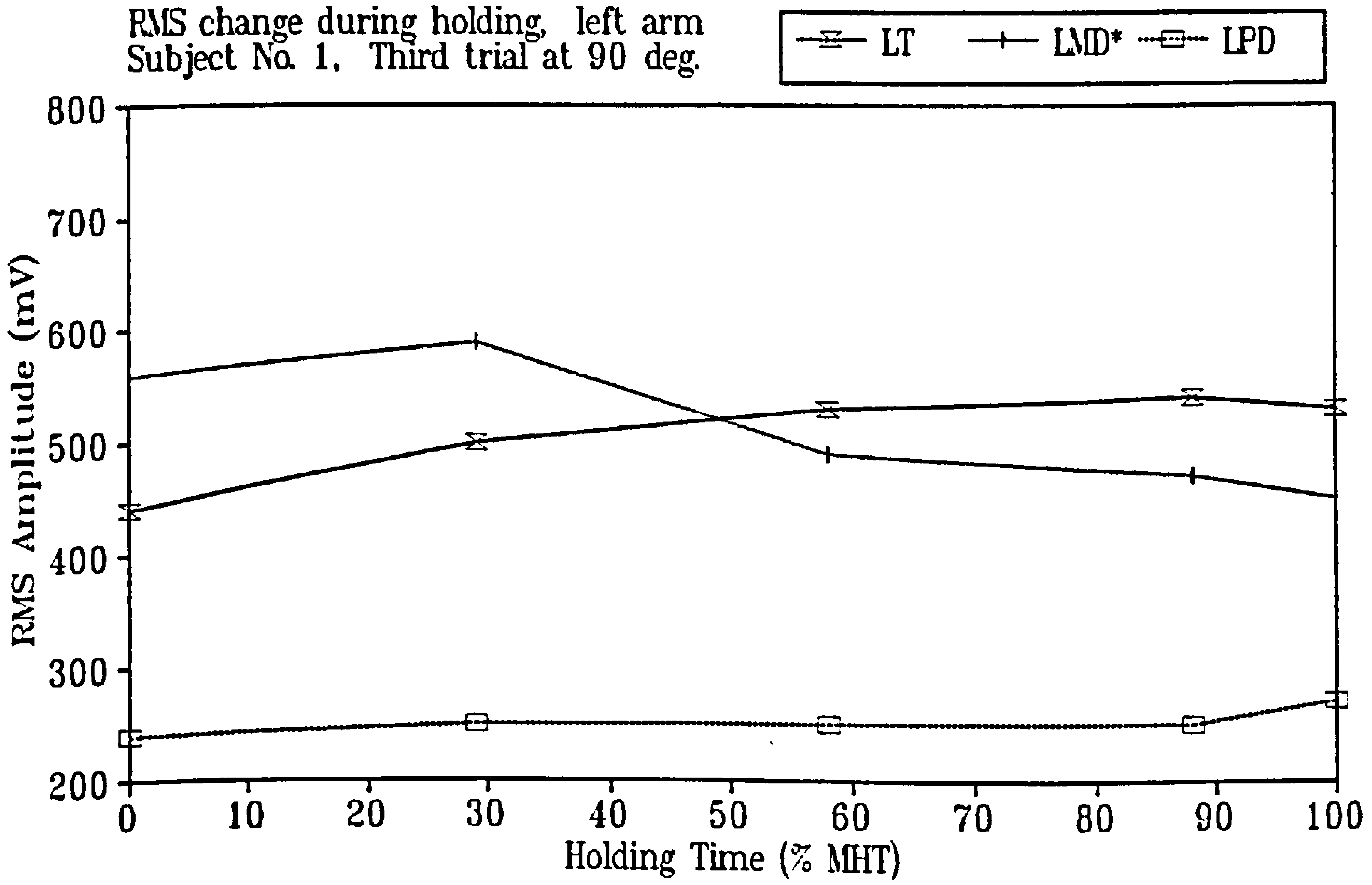


Fig D.21 Change in RMS amplitude for subject No. 1 at 90 deg.

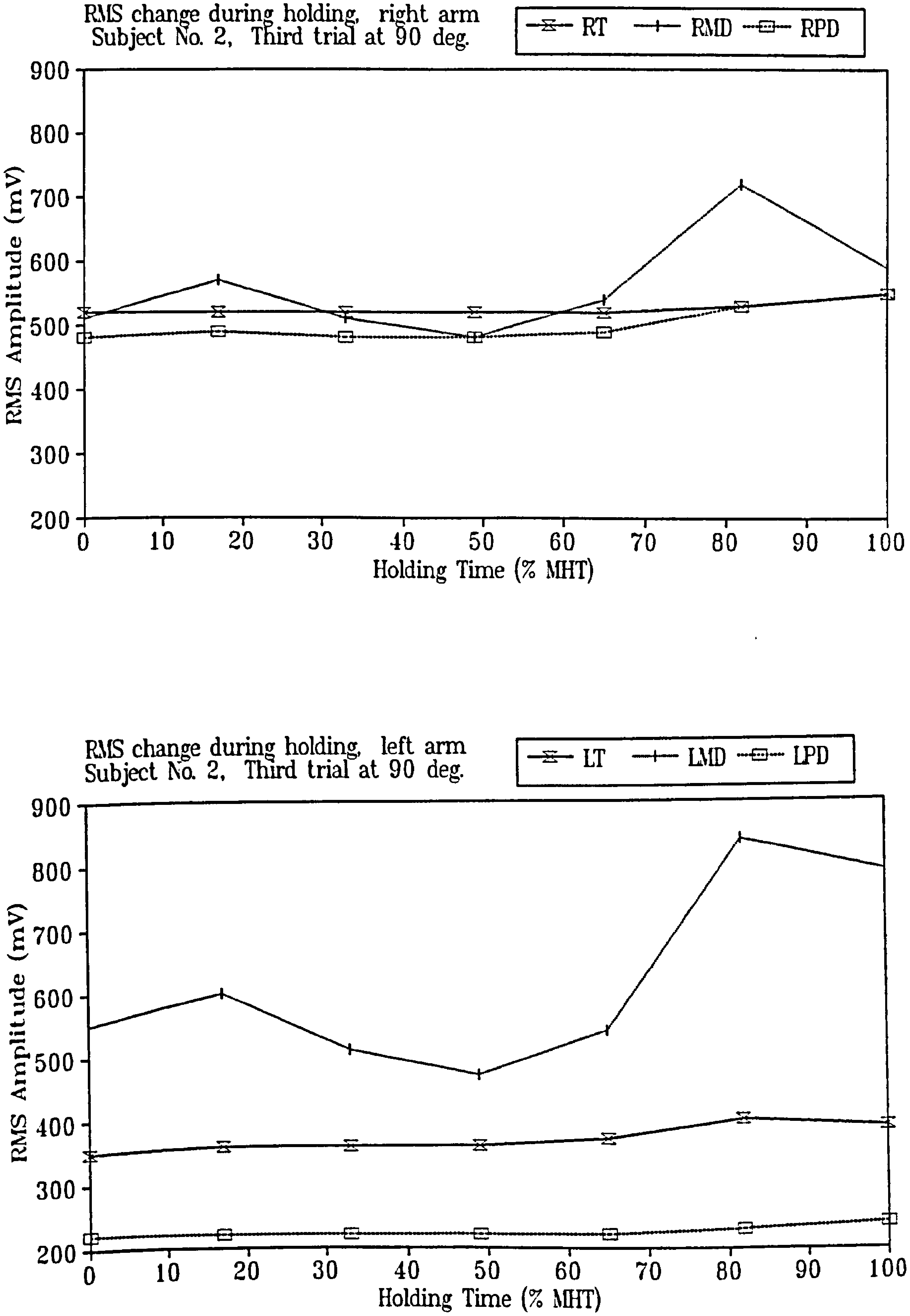
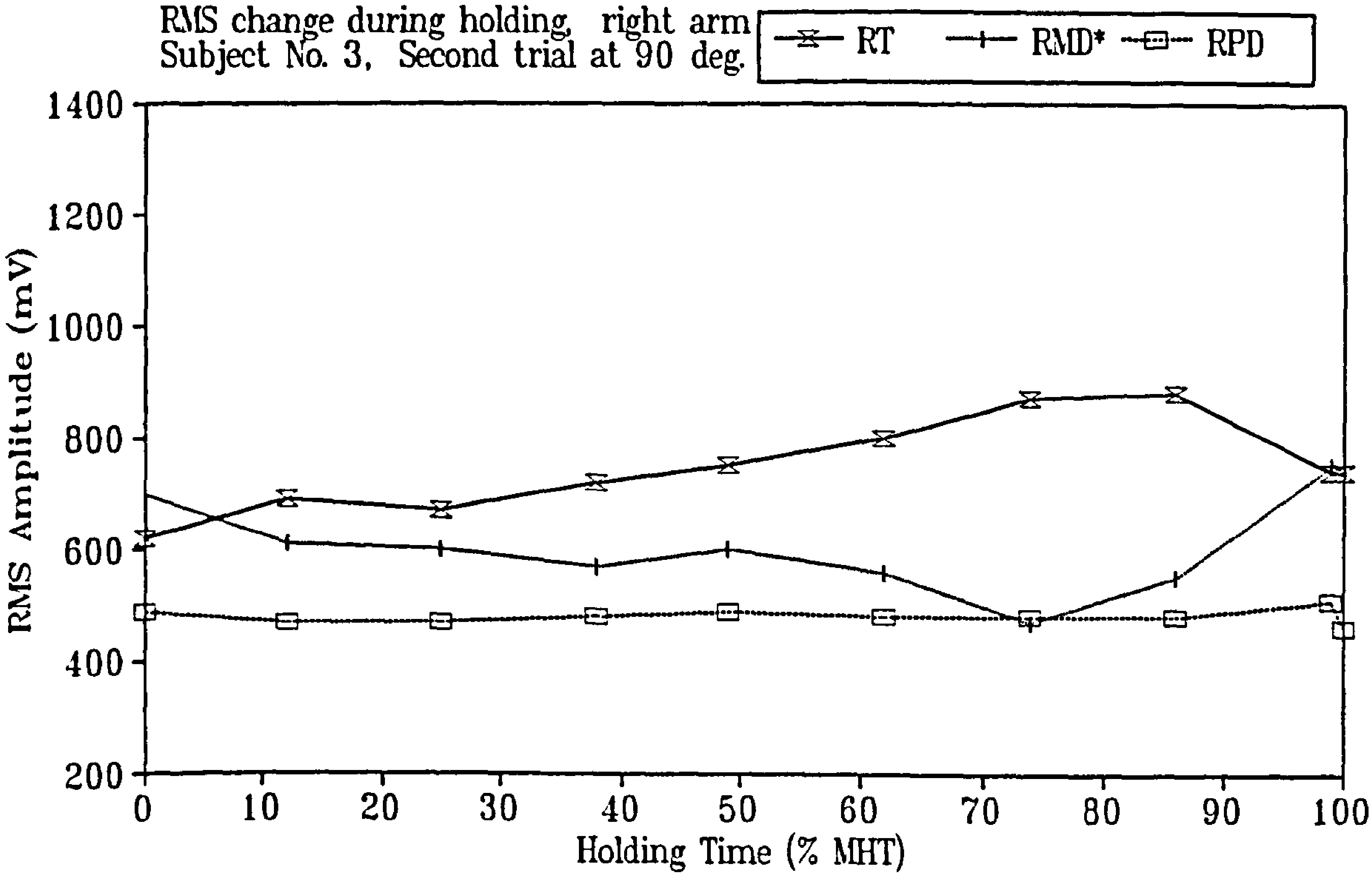


Fig D.22 Changes in RMS amplitude for subject No. 2 at 90 deg.



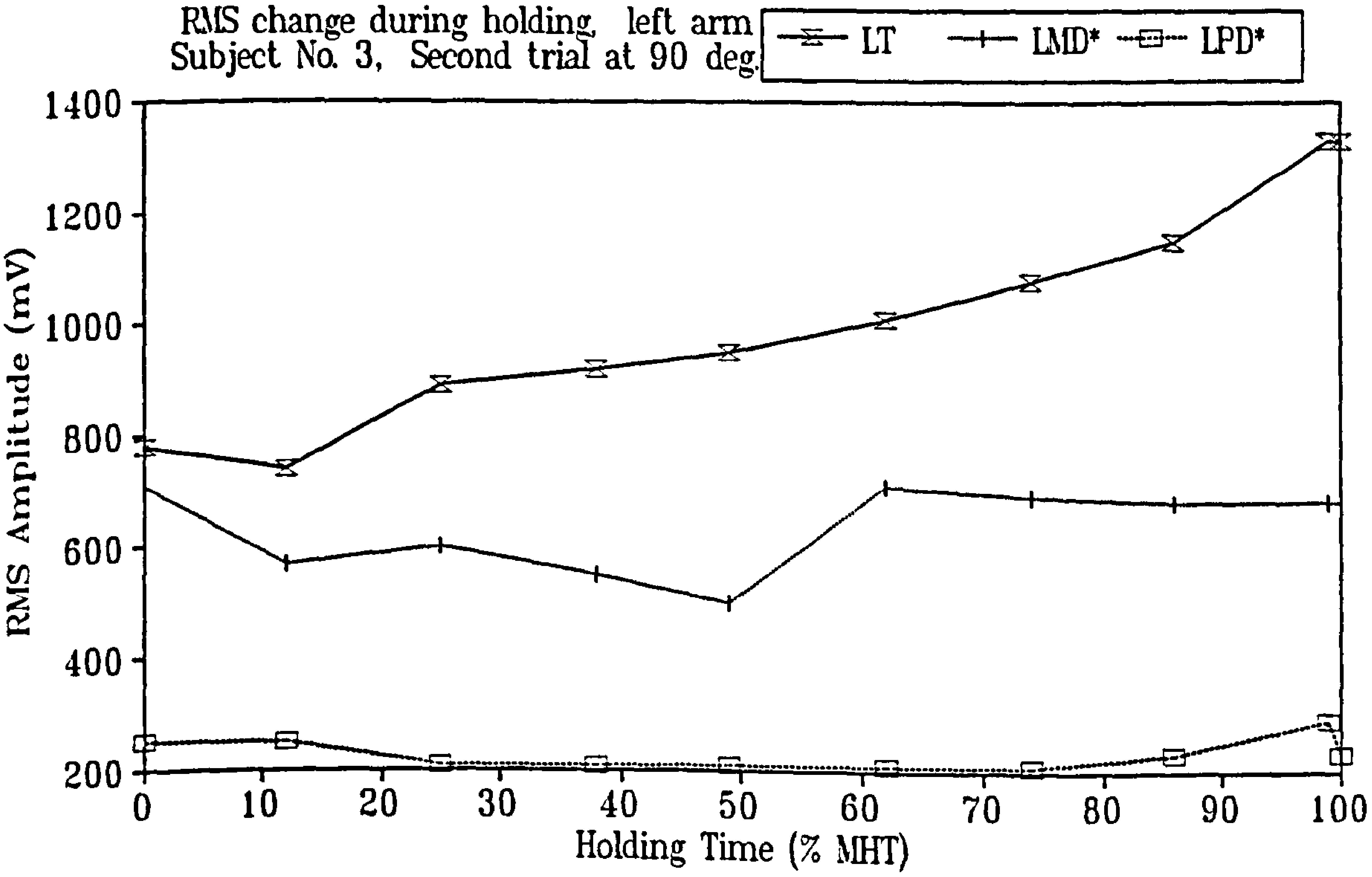
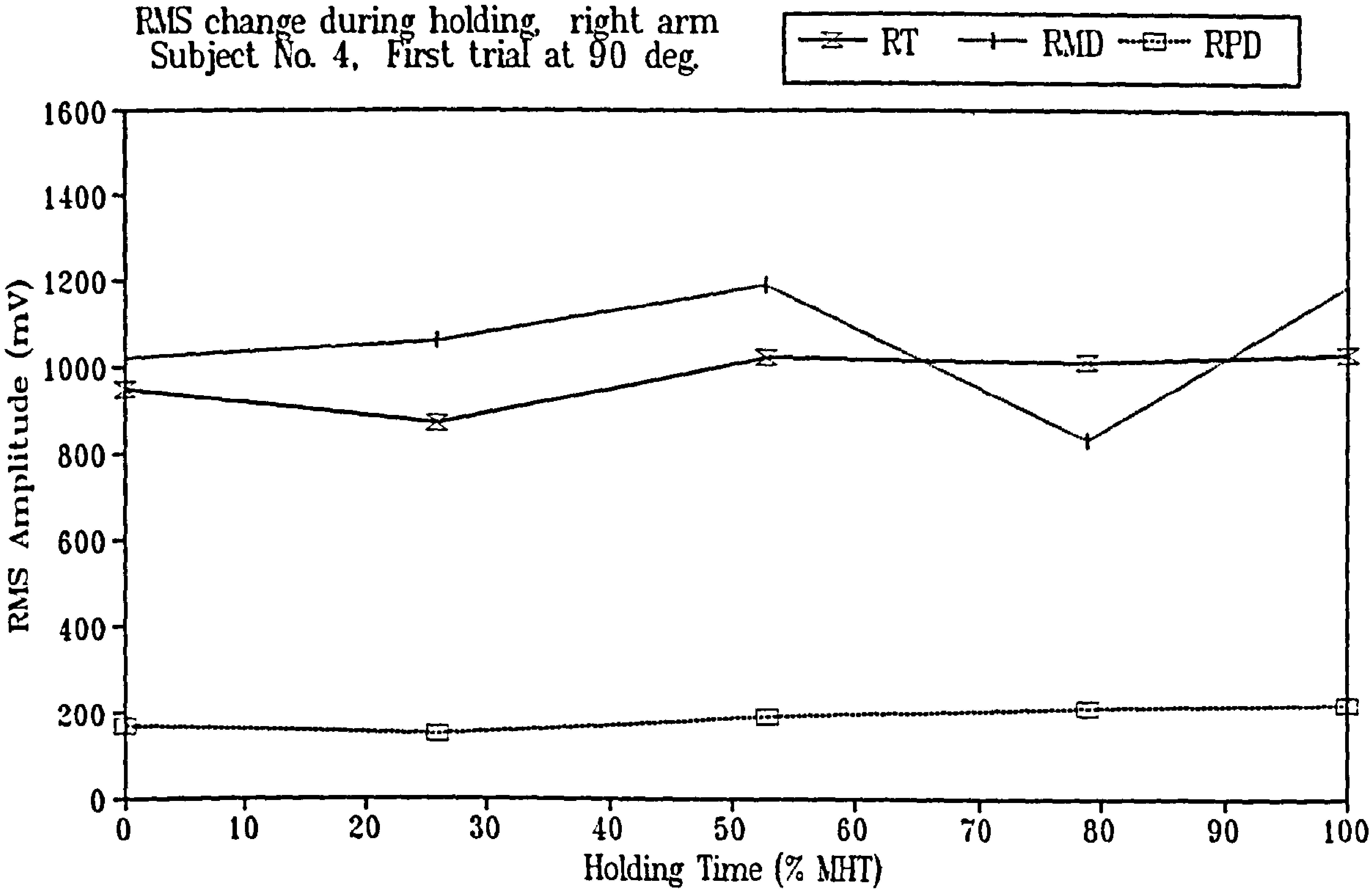


Fig D.23 Changes in RMS amplitude for subject No. 3 at 90 deg.



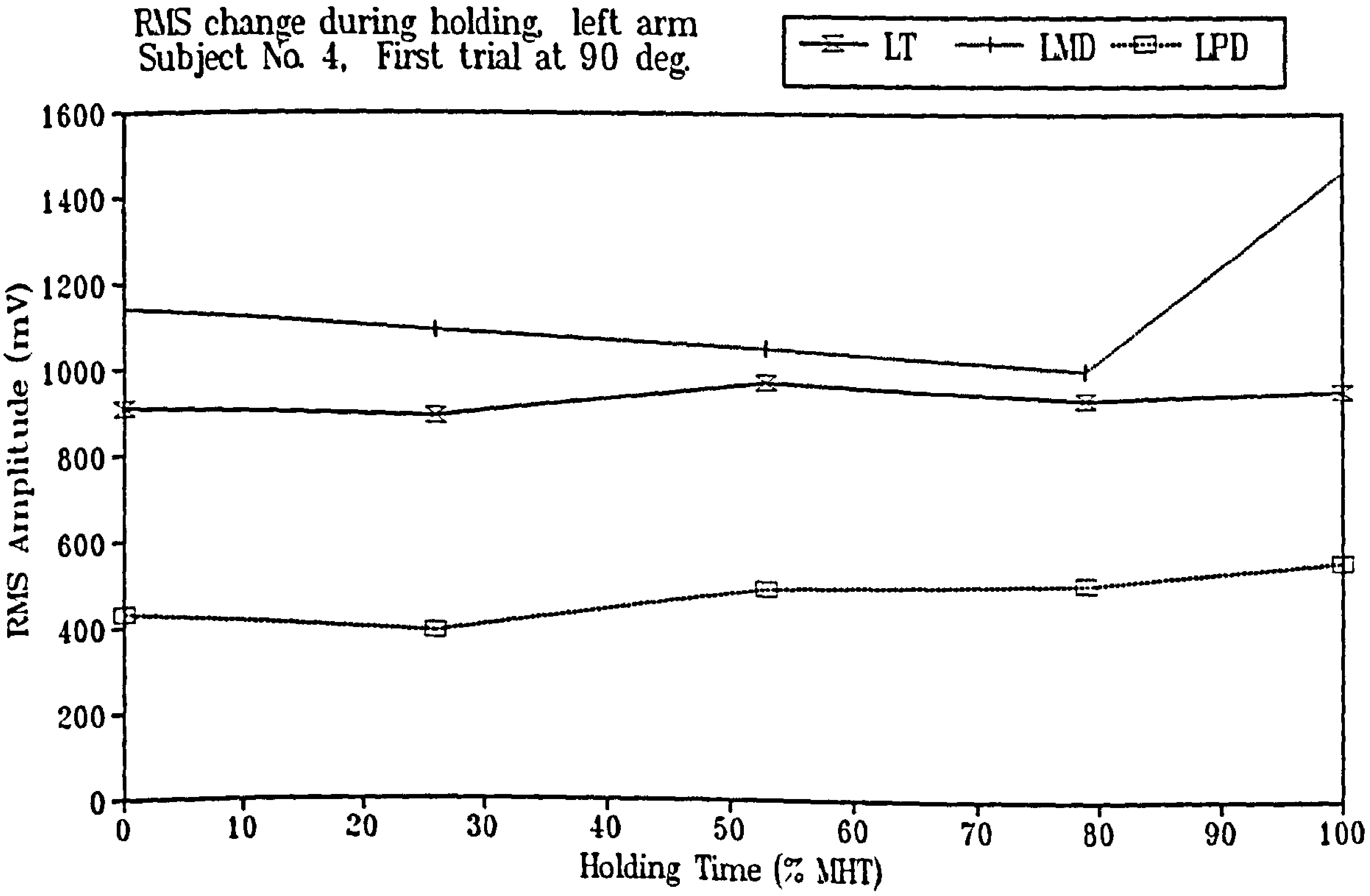
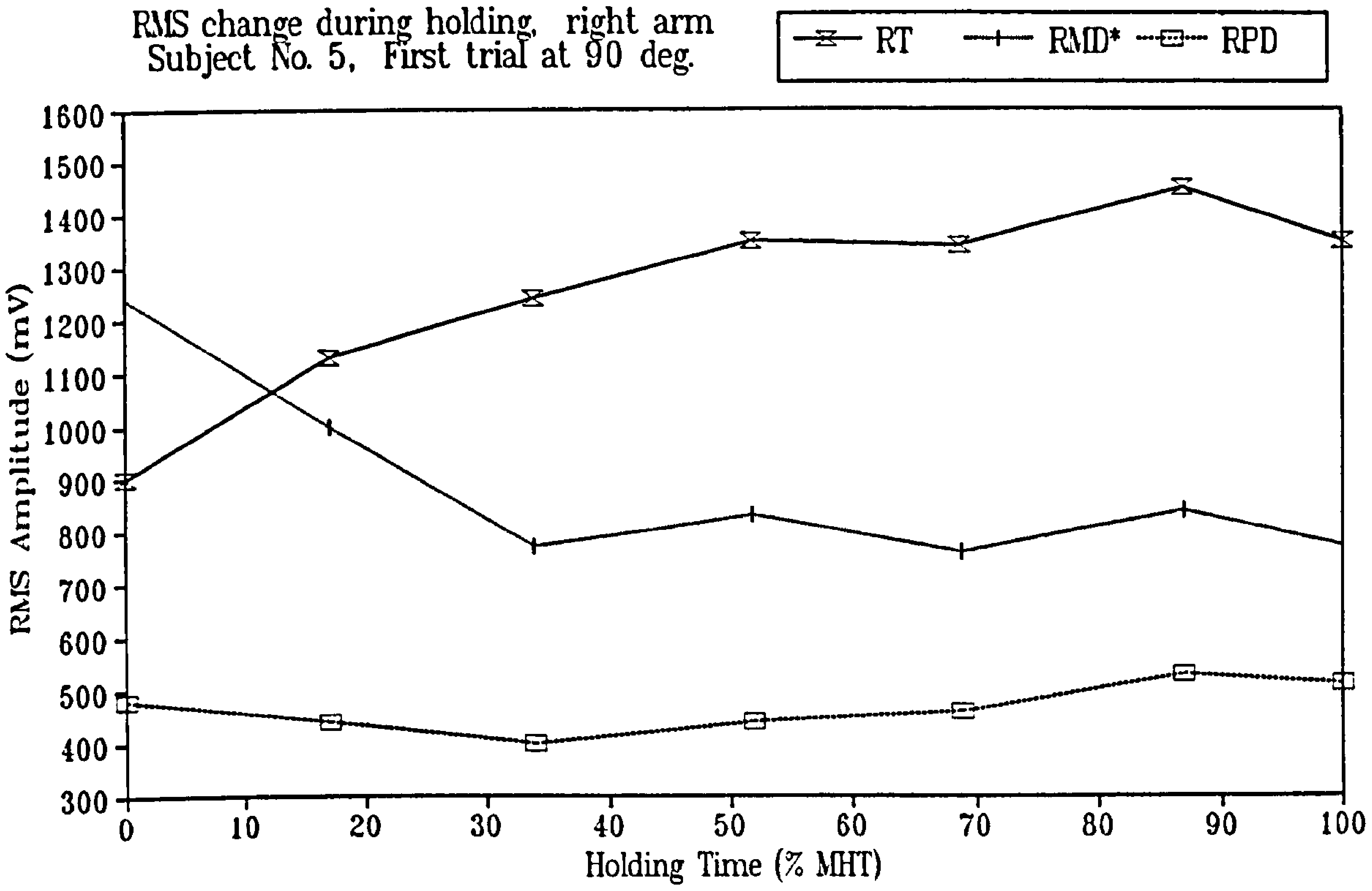


Figure D.24 Changes in RMS amplitude for subject No. 4 at 90 deg.



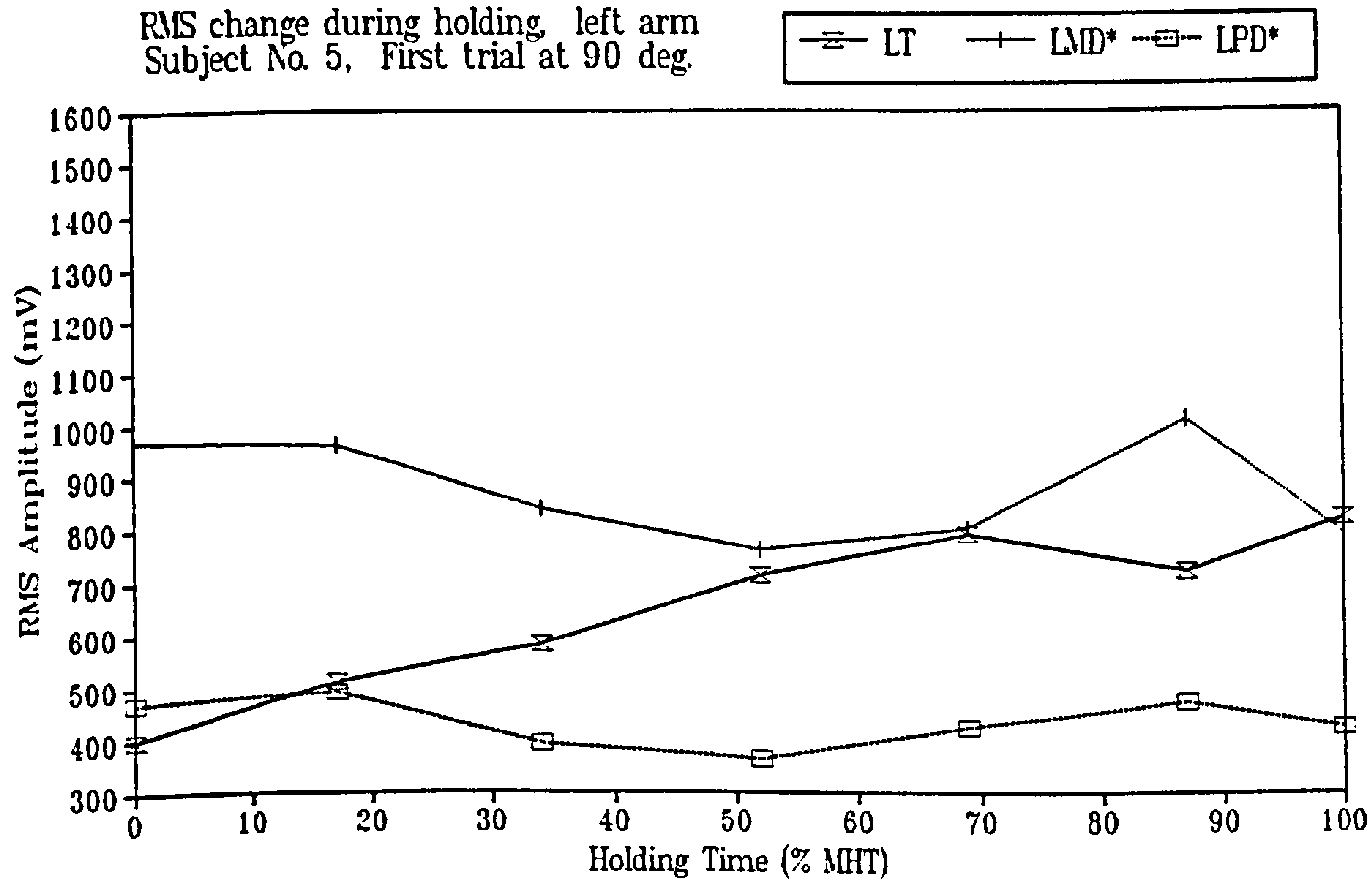
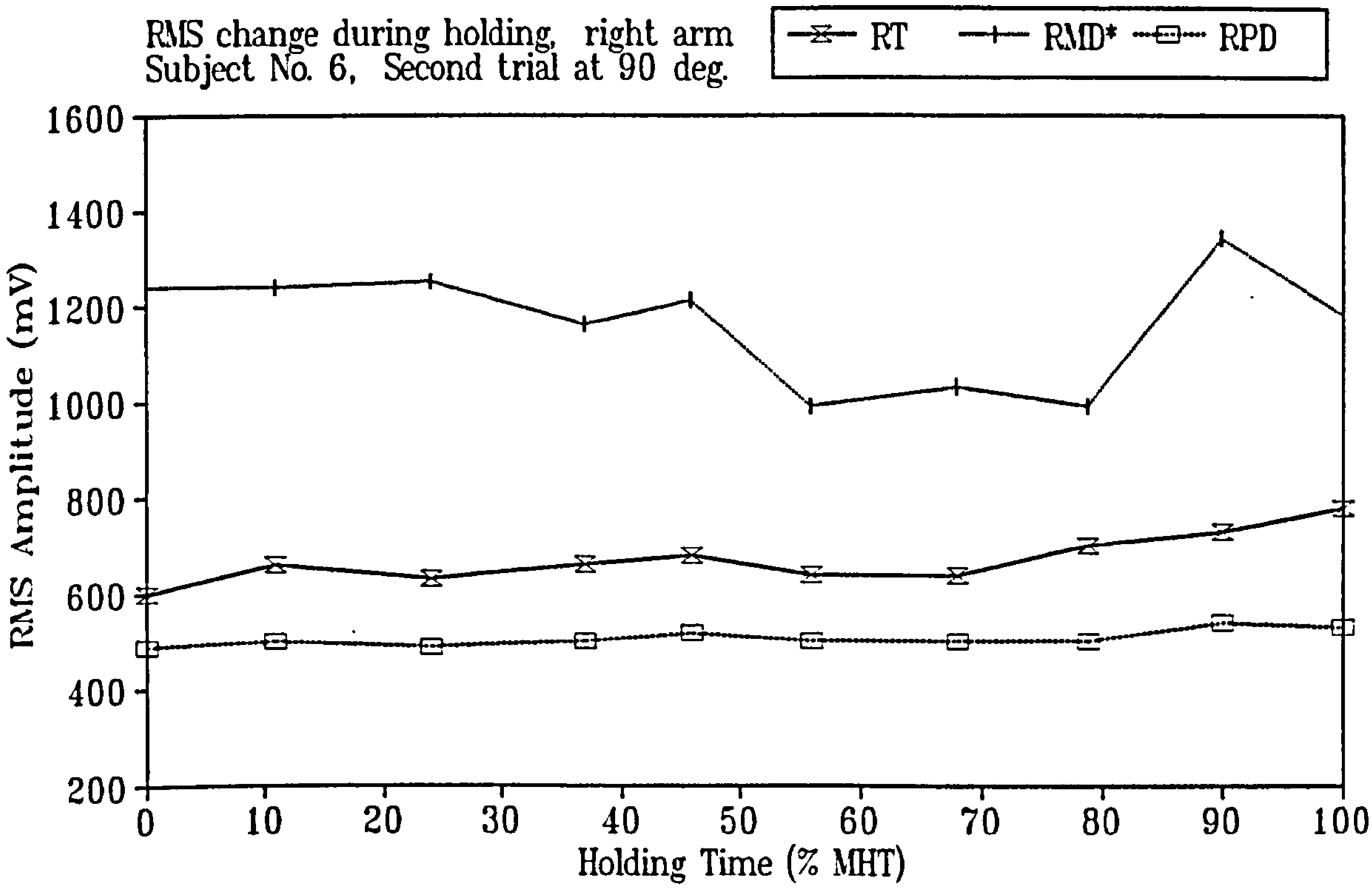


Figure D.25 Changes in RMS amplitude for subject No. 5 at 90 deg.



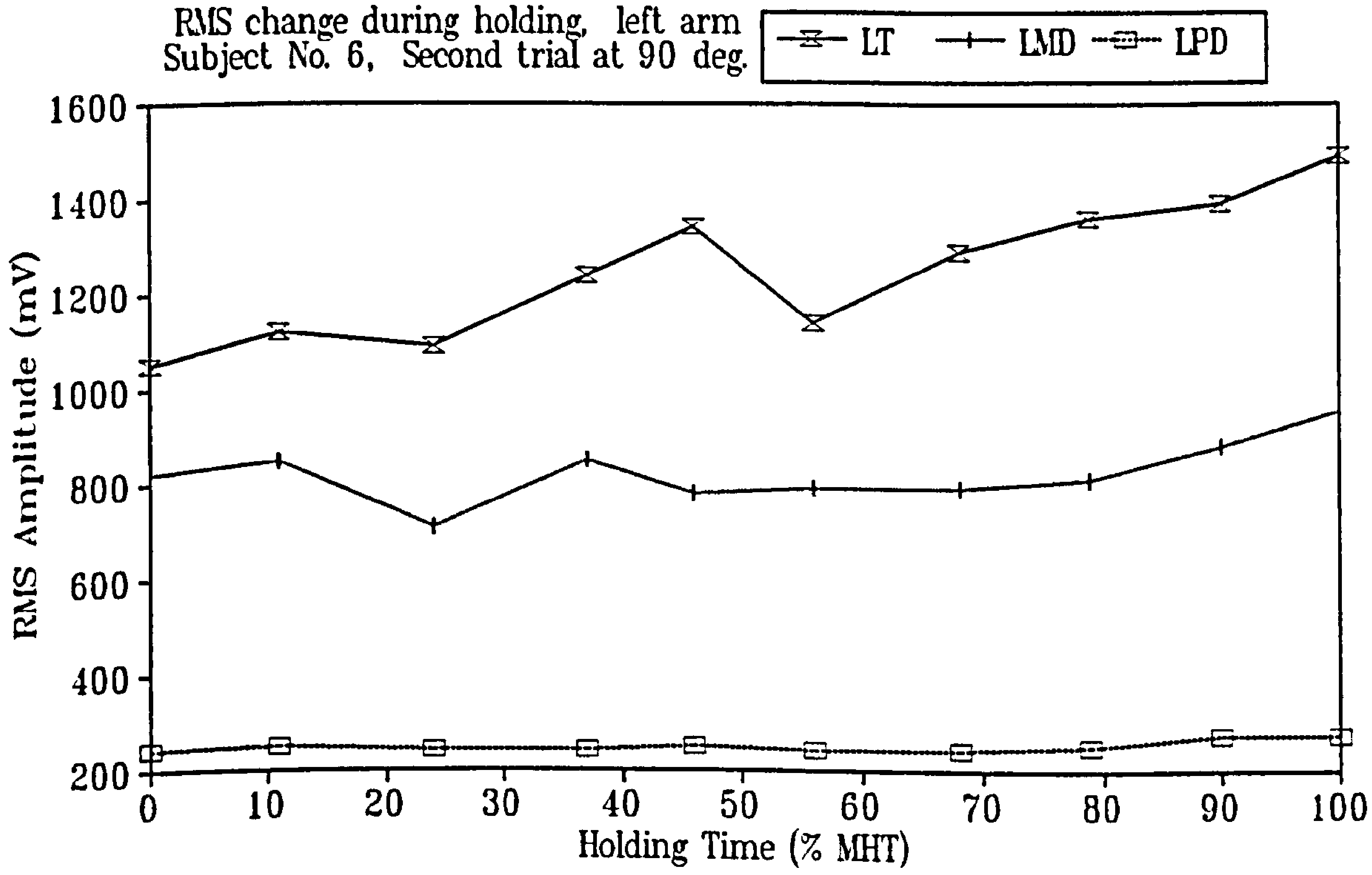
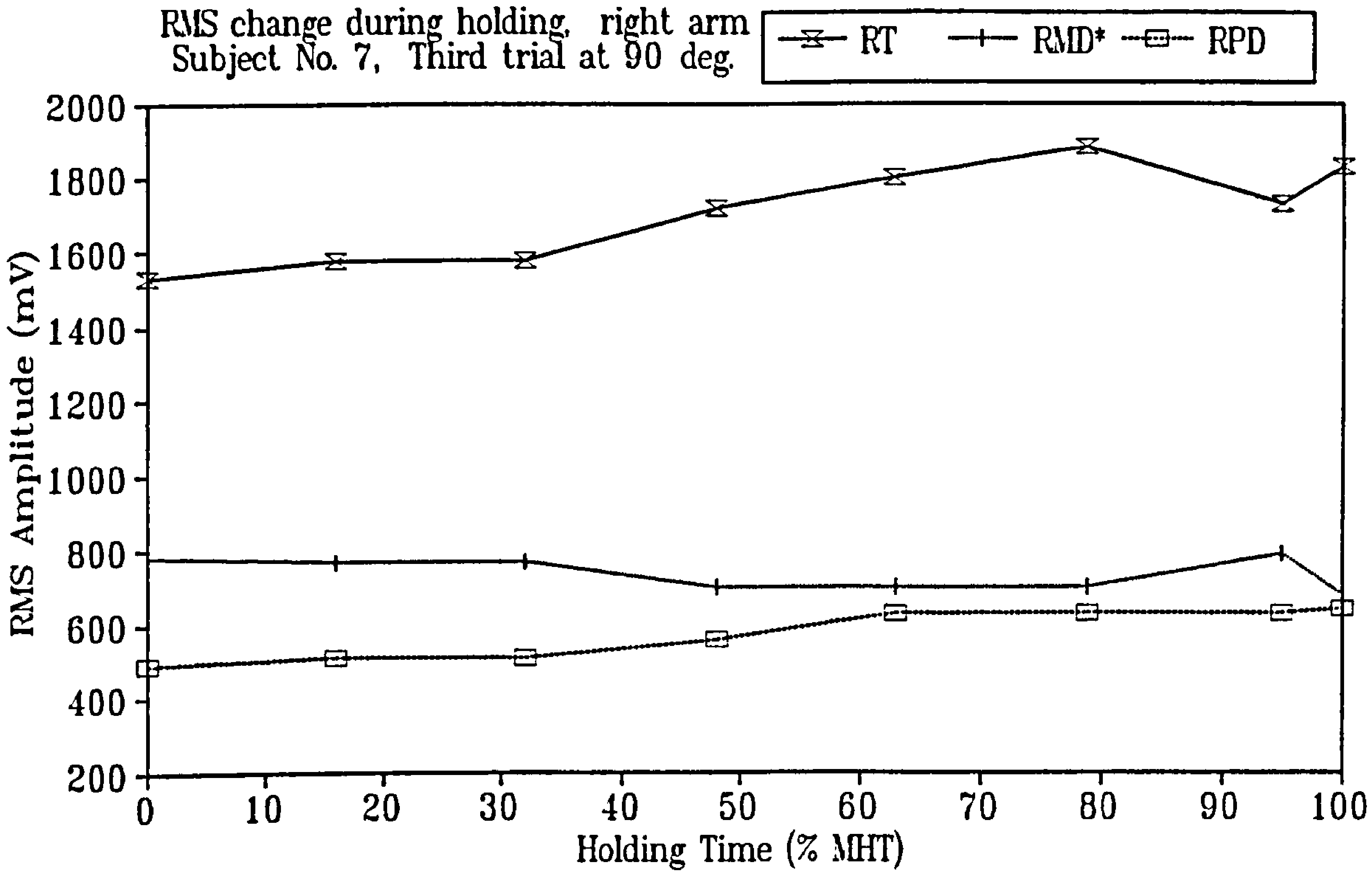


Figure D.26 Changes in RMS amplitude for subject No. 6 at 90 deg.



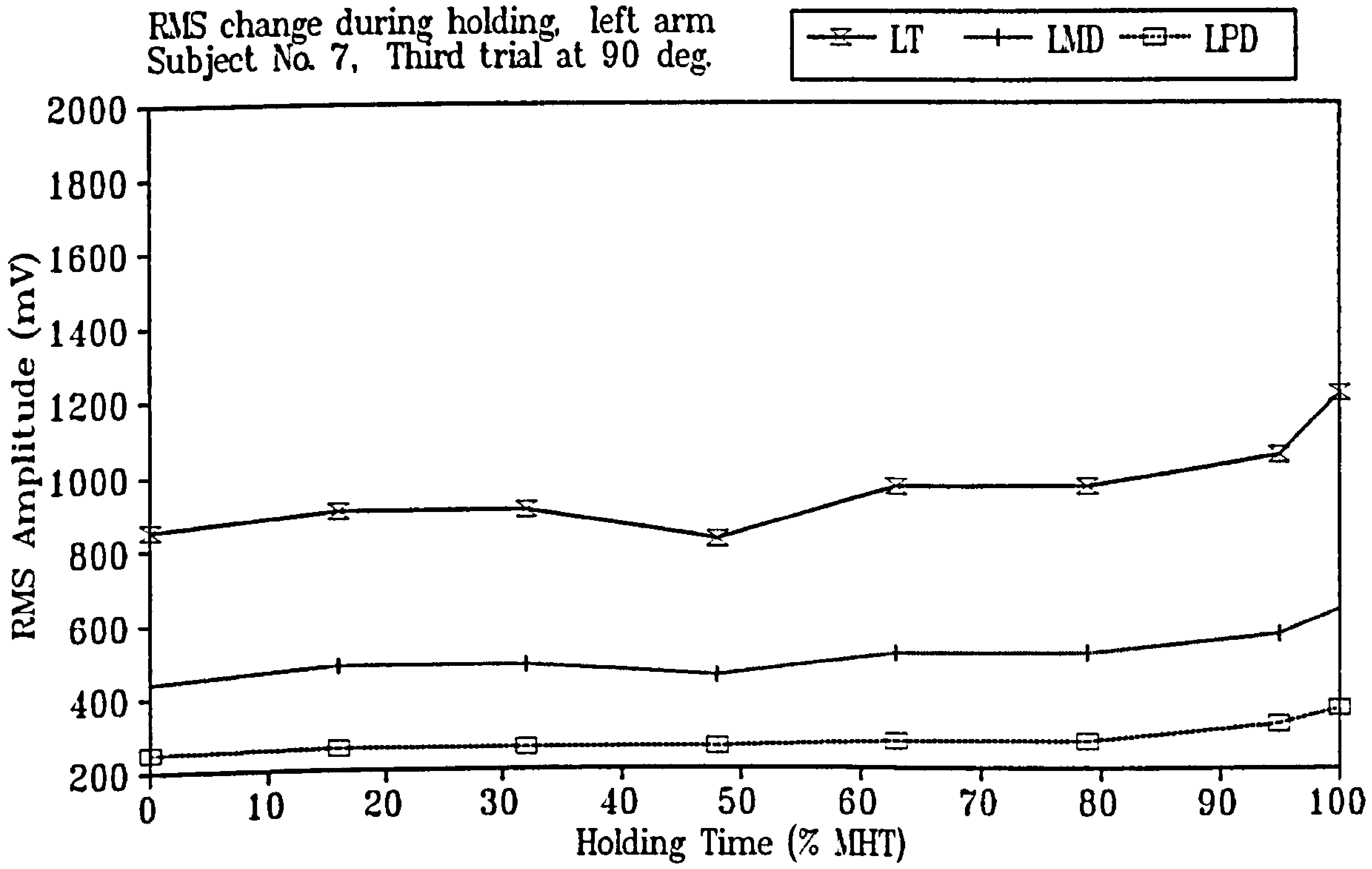
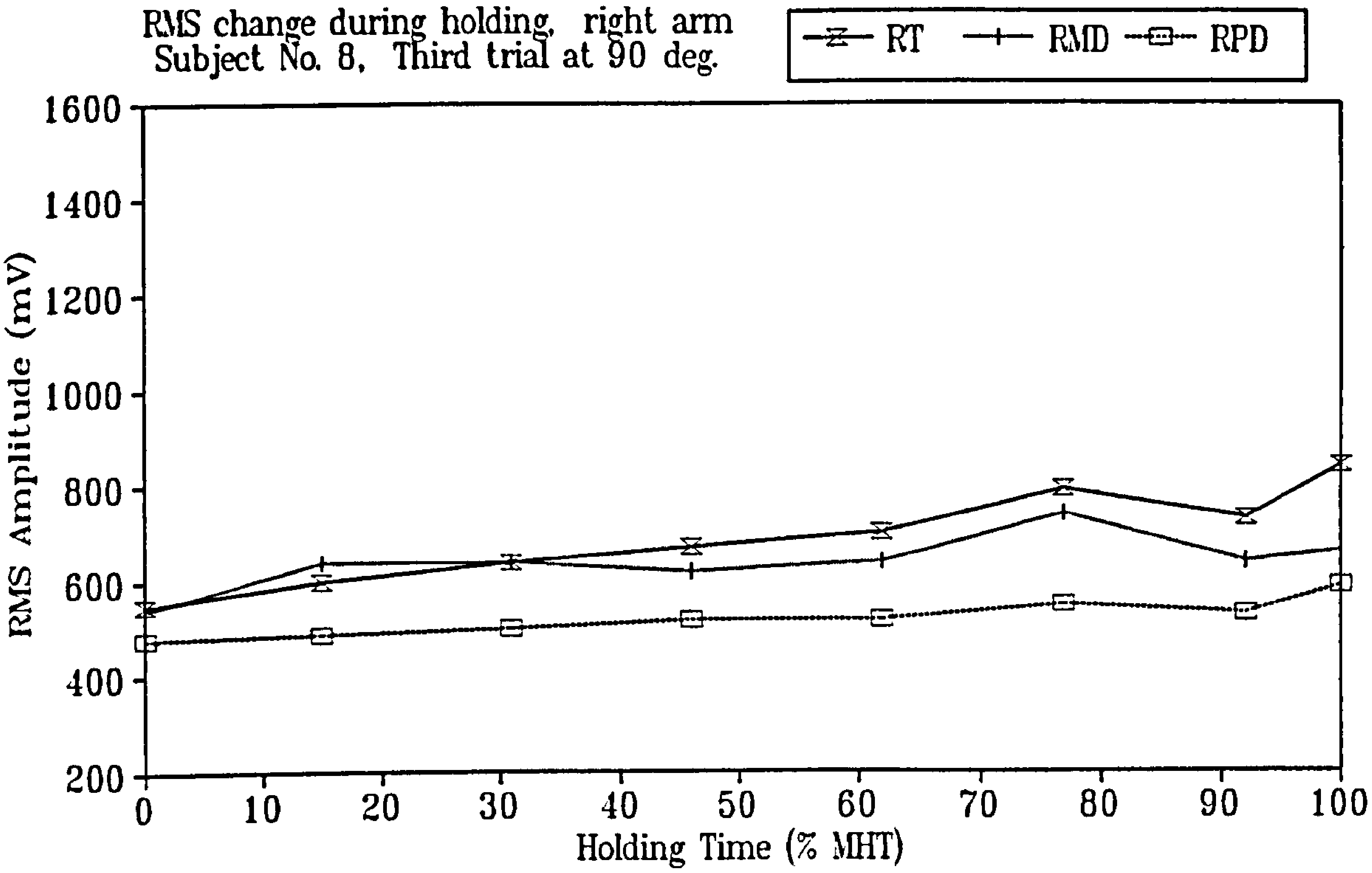


Fig D.27 Changes in RMS amplitude for subject No. 7 at 90 deg.



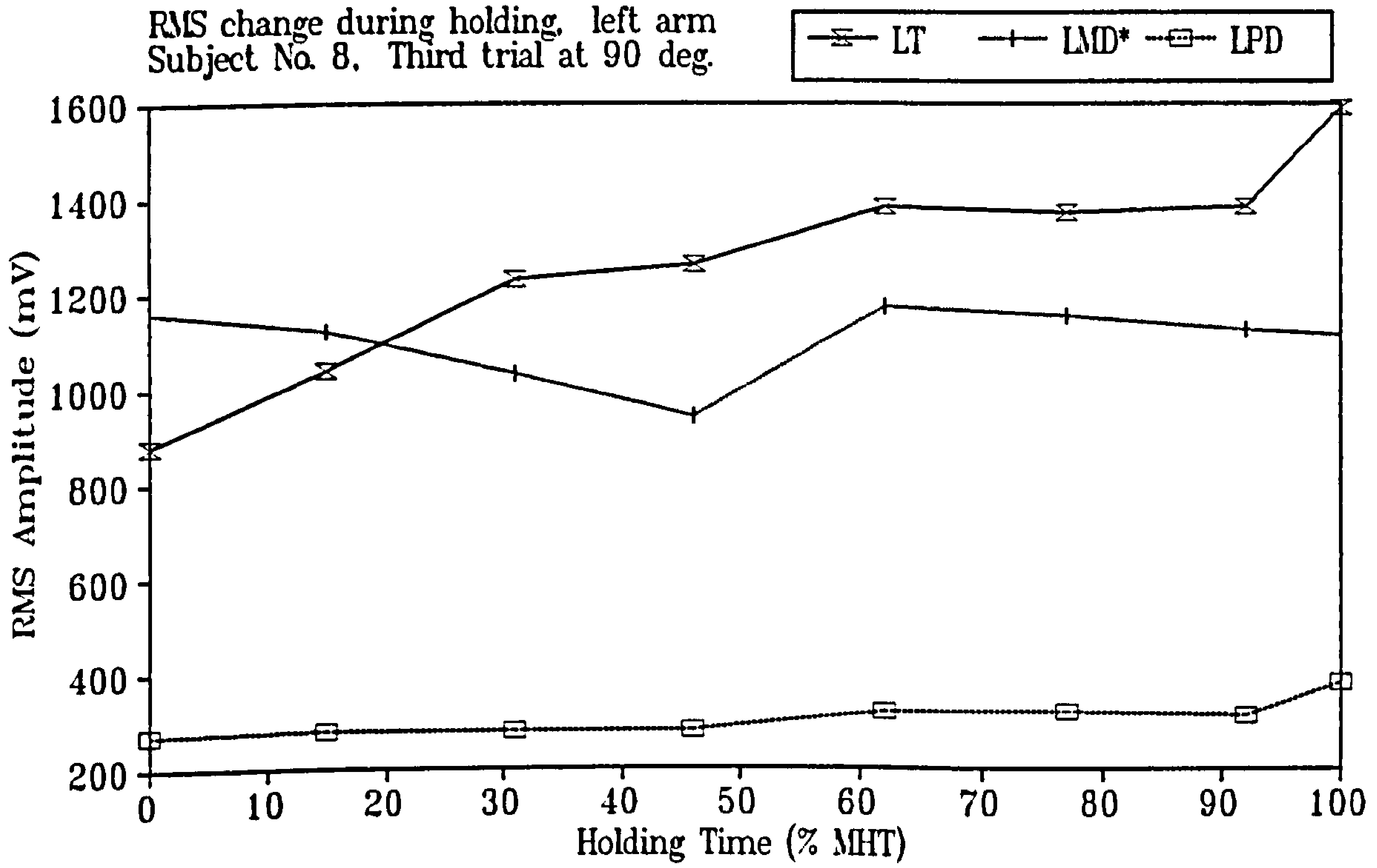
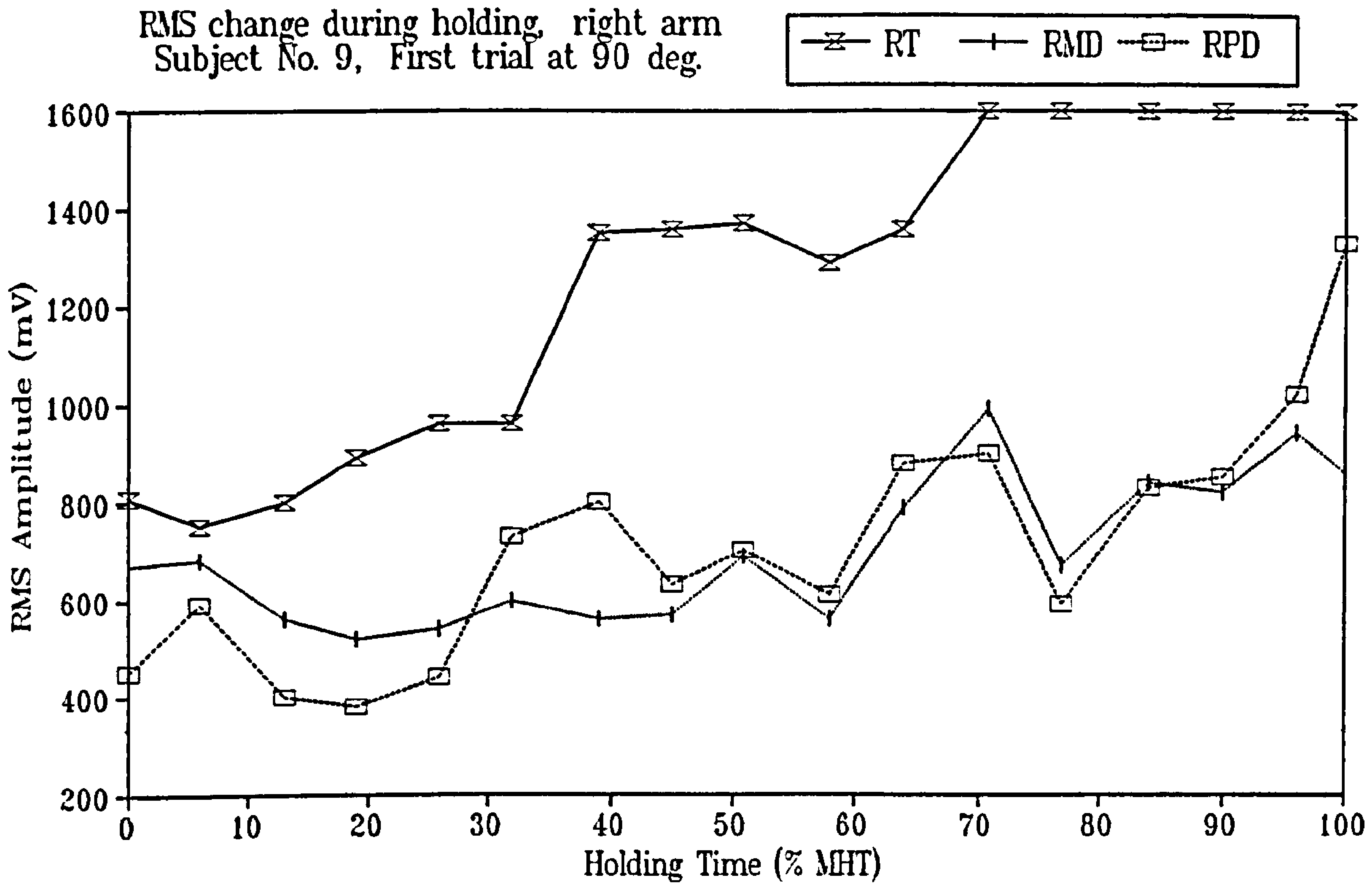


Figure D.28 Changes in RMS amplitude for subject No. 8 at 90 deg.



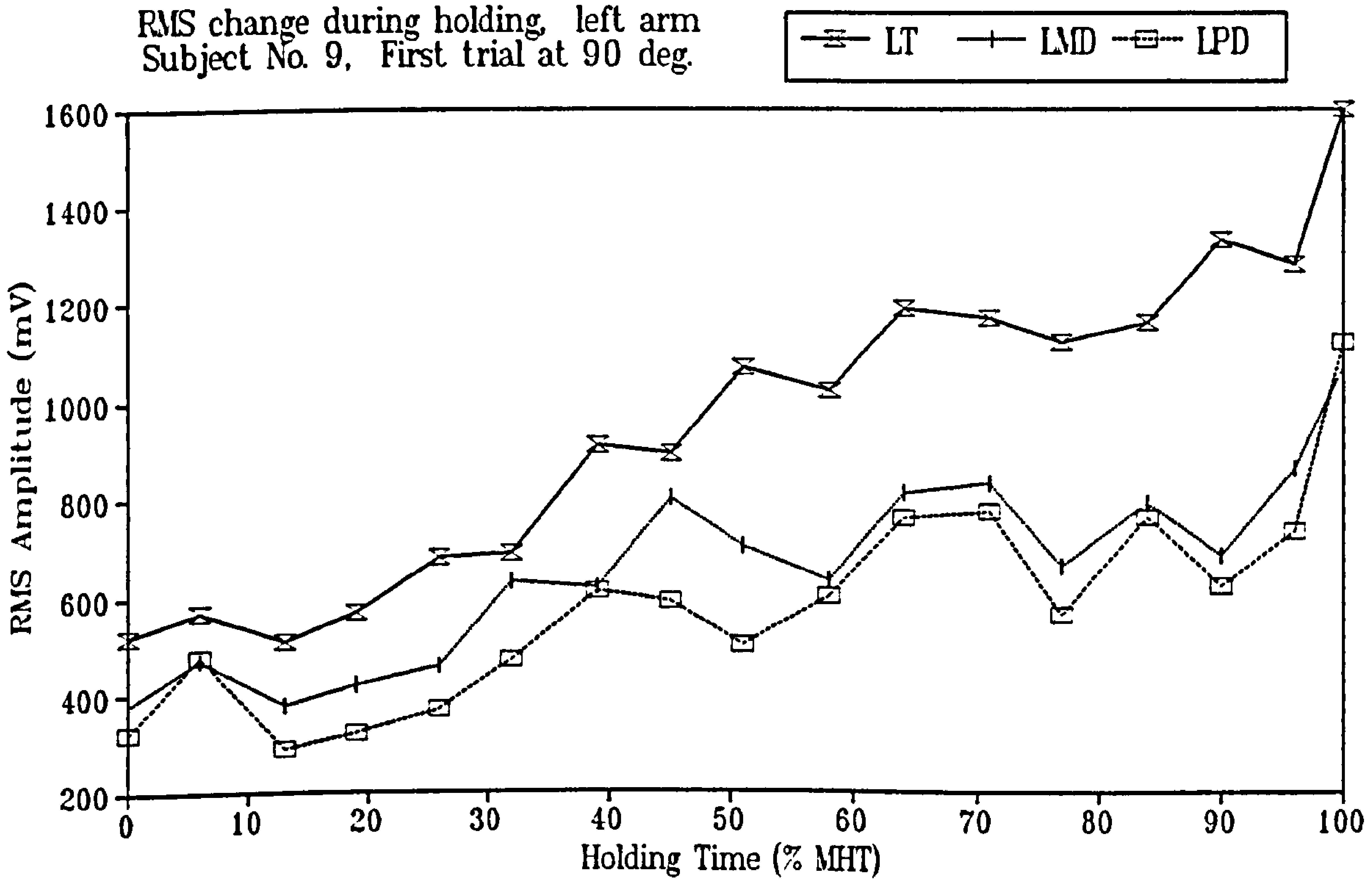
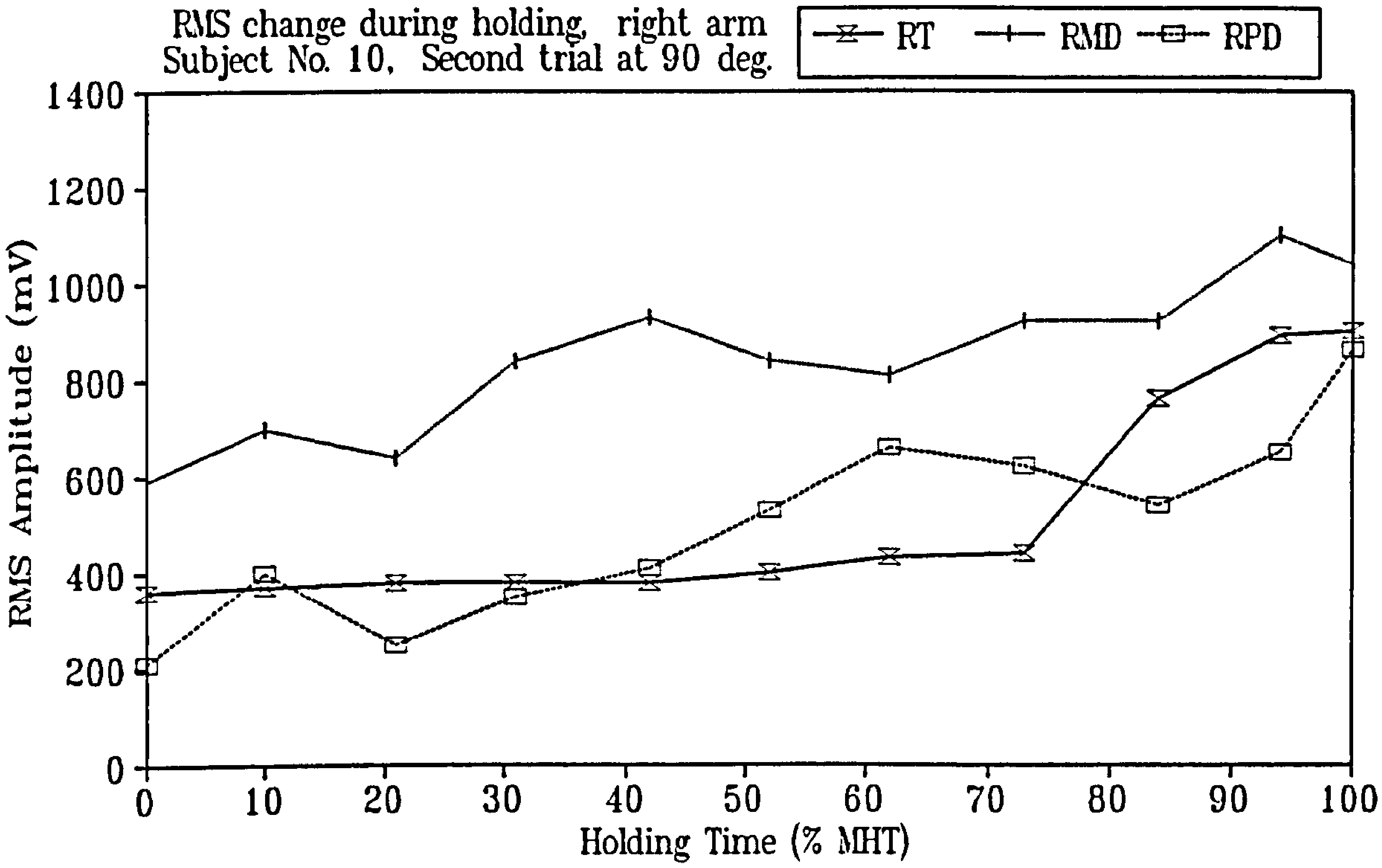


Figure D.29 Changes in RMS amplitude for subject No. 9 at 90 deg.



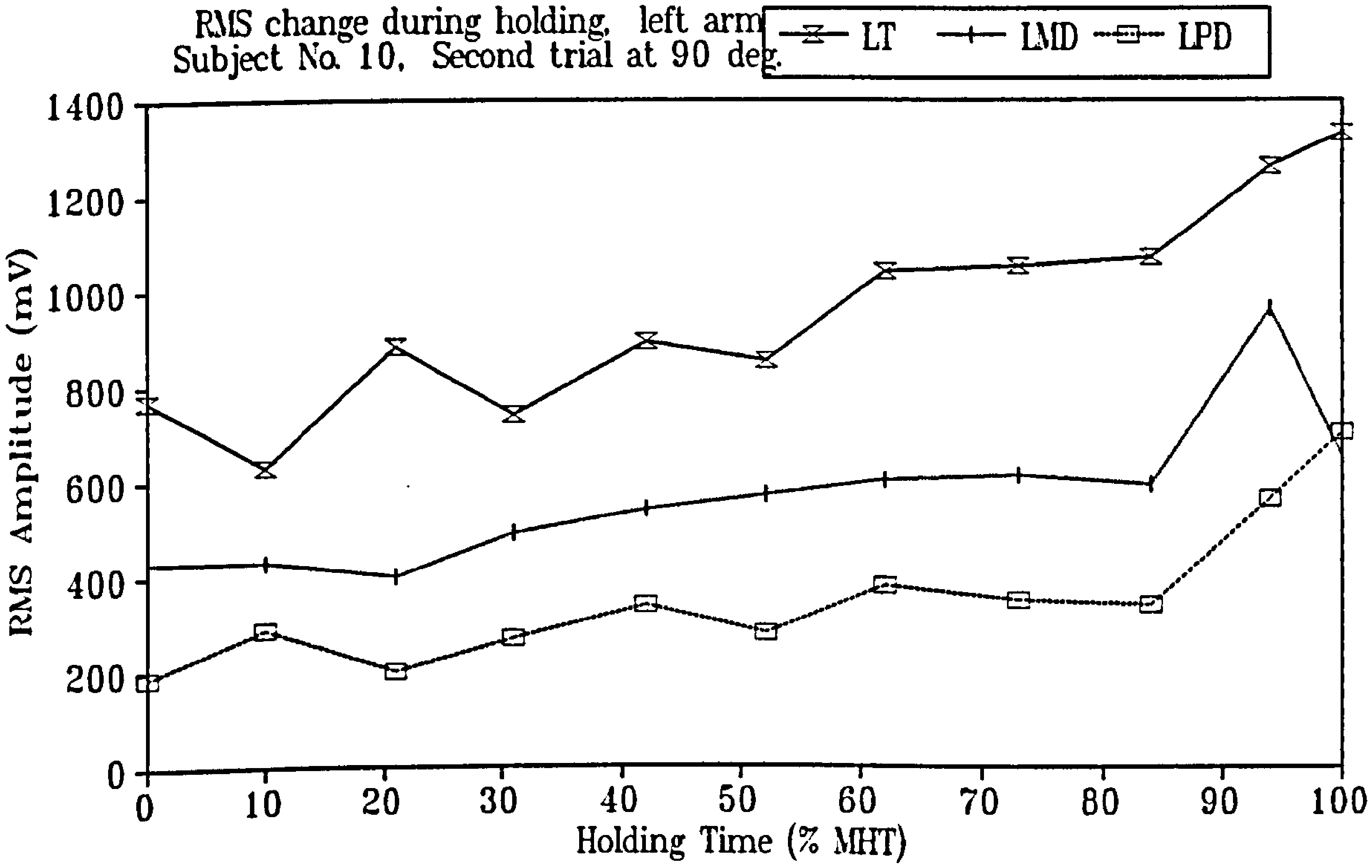
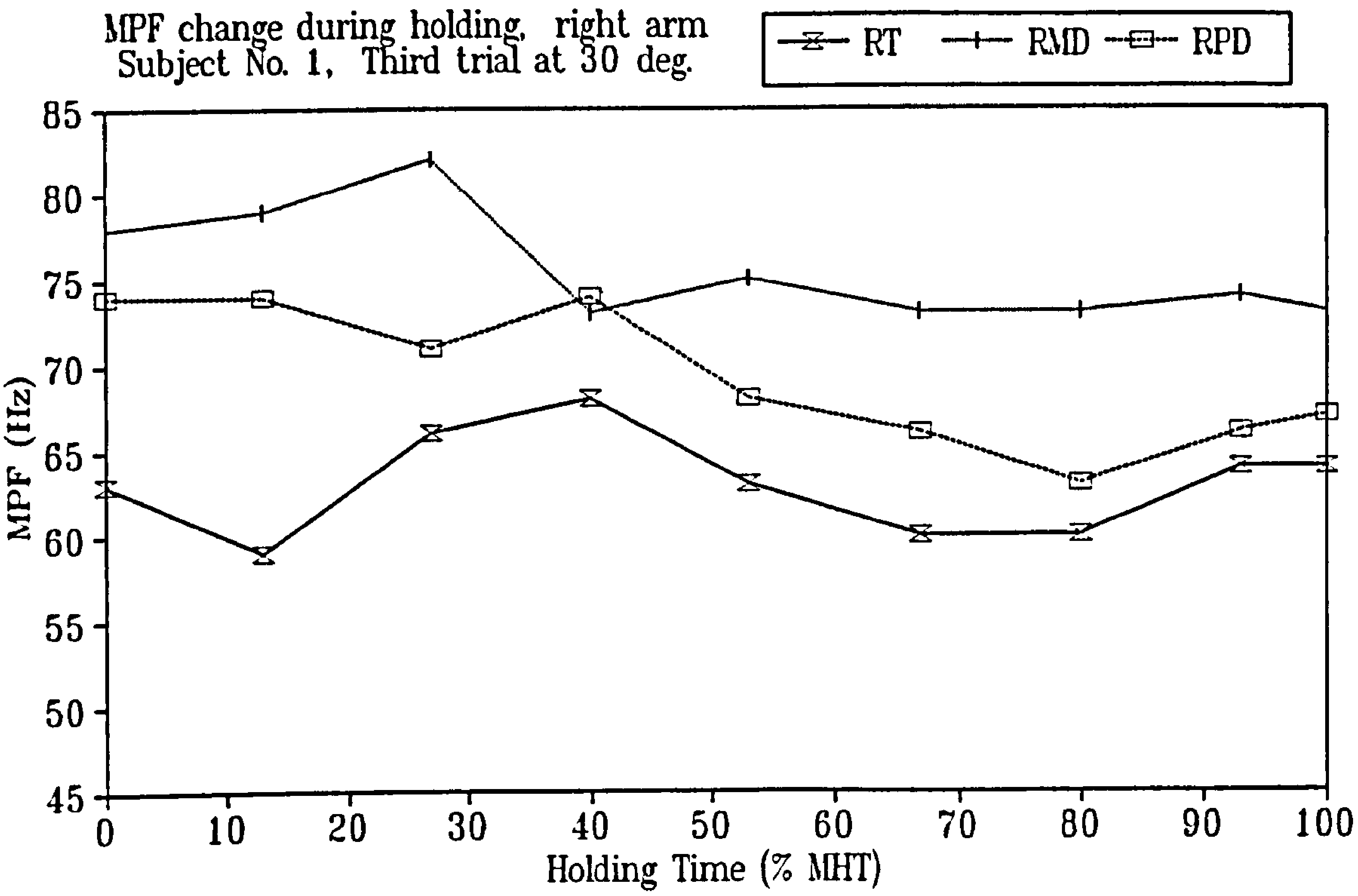


Figure D.30 Changes in RMS amplitude for subject No. 10 at 90 deg.



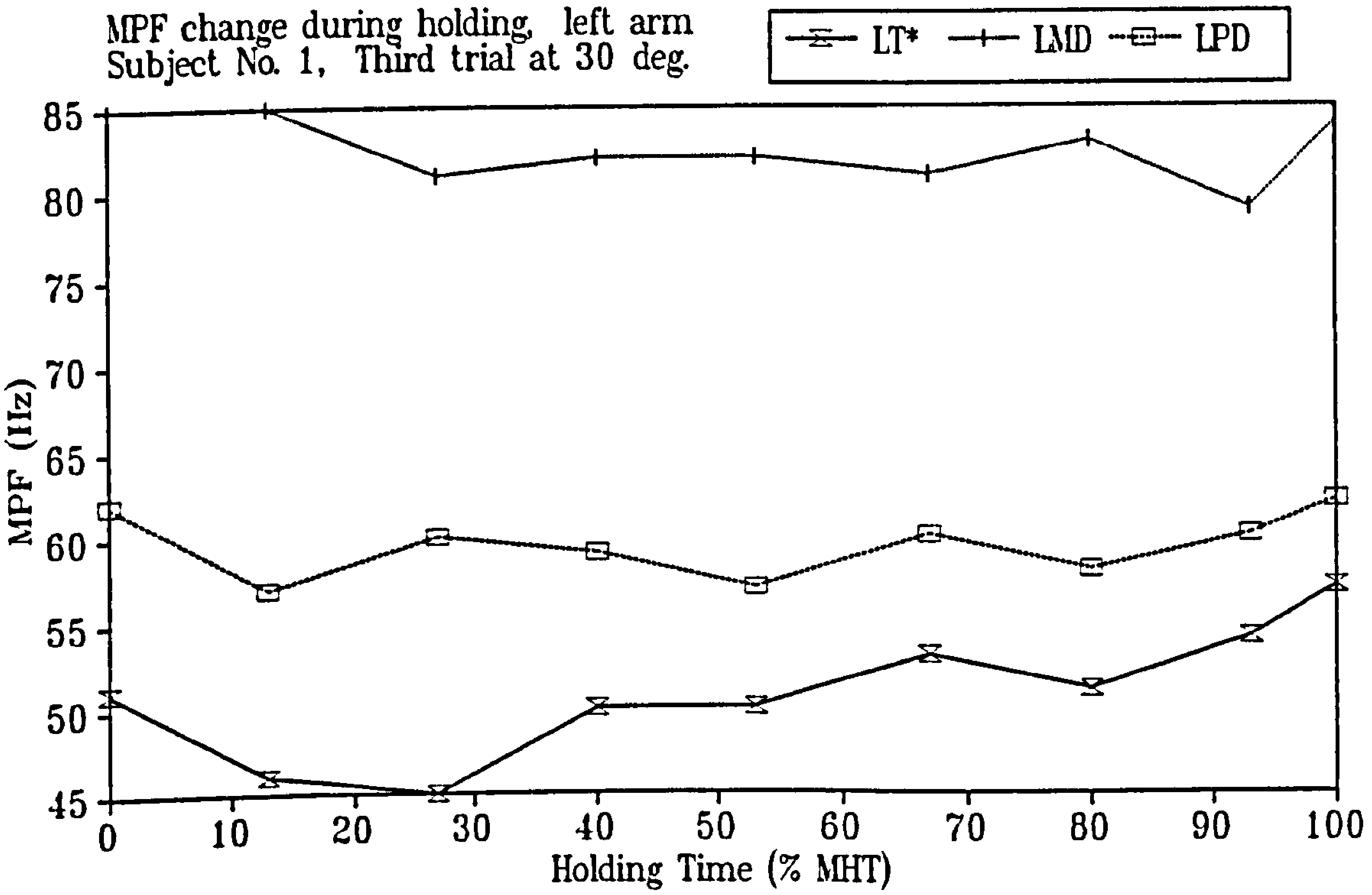
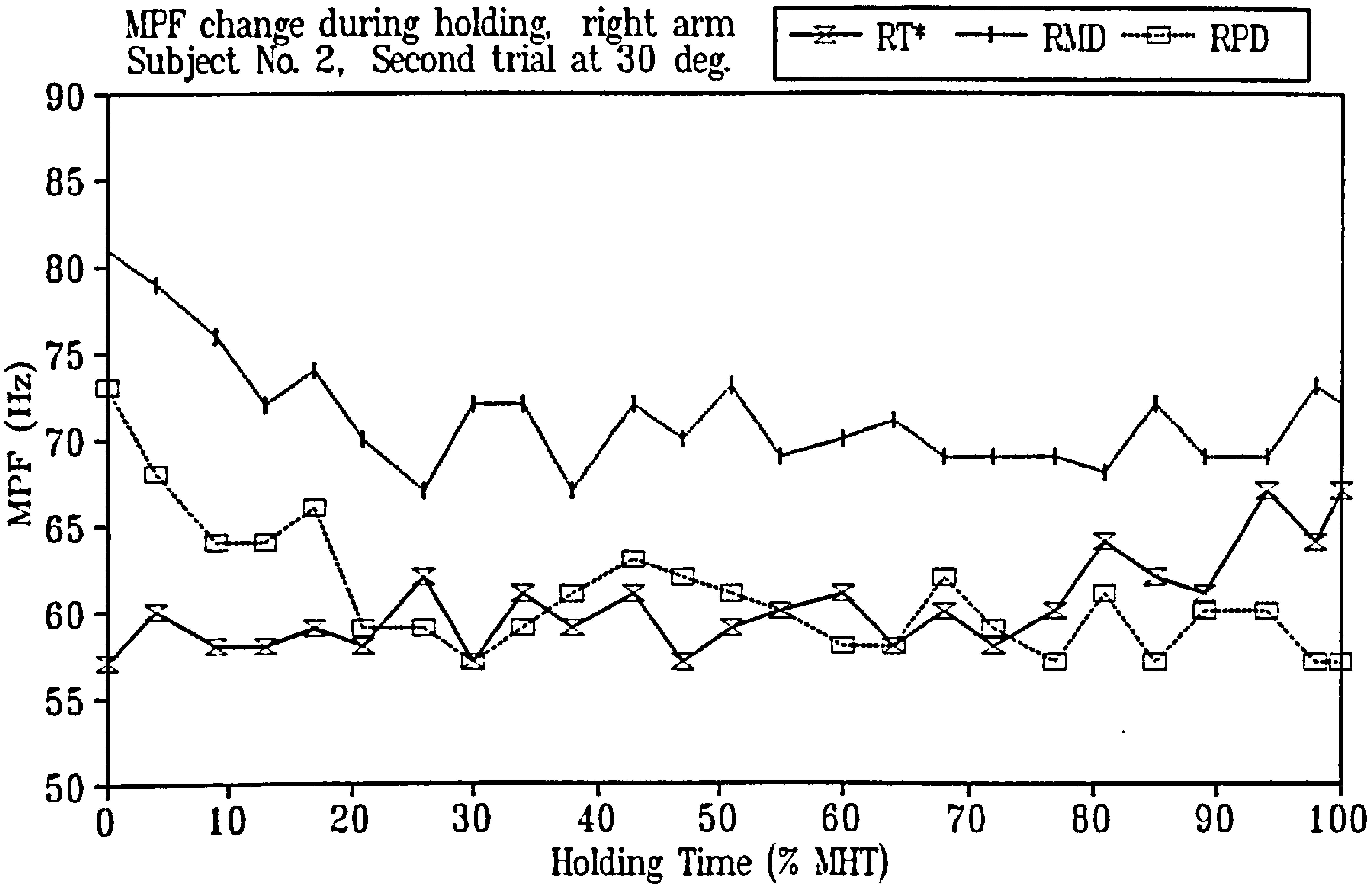


Figure D.31 Changes in MPF for subject No. 1 at 30 deg.



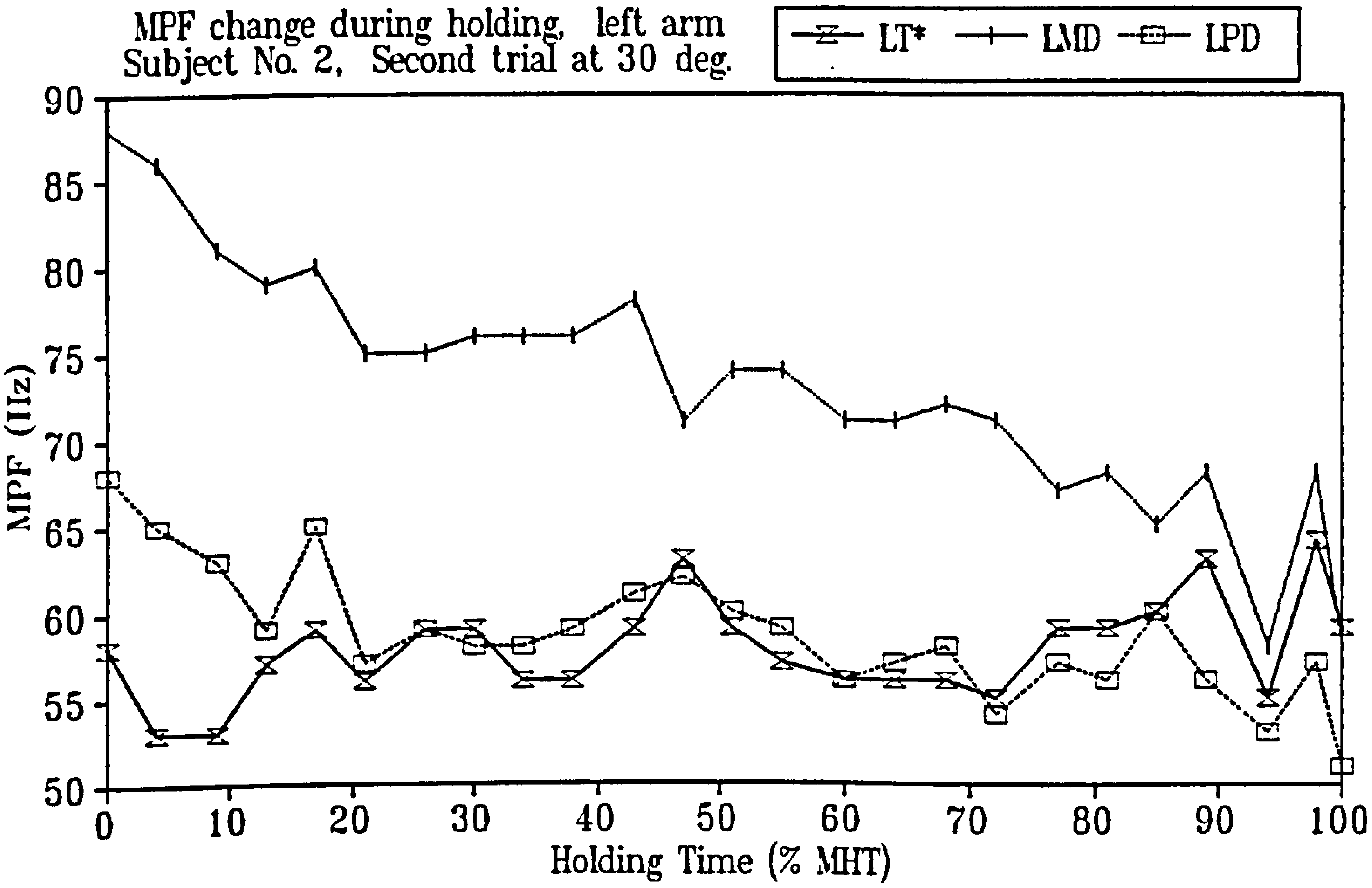
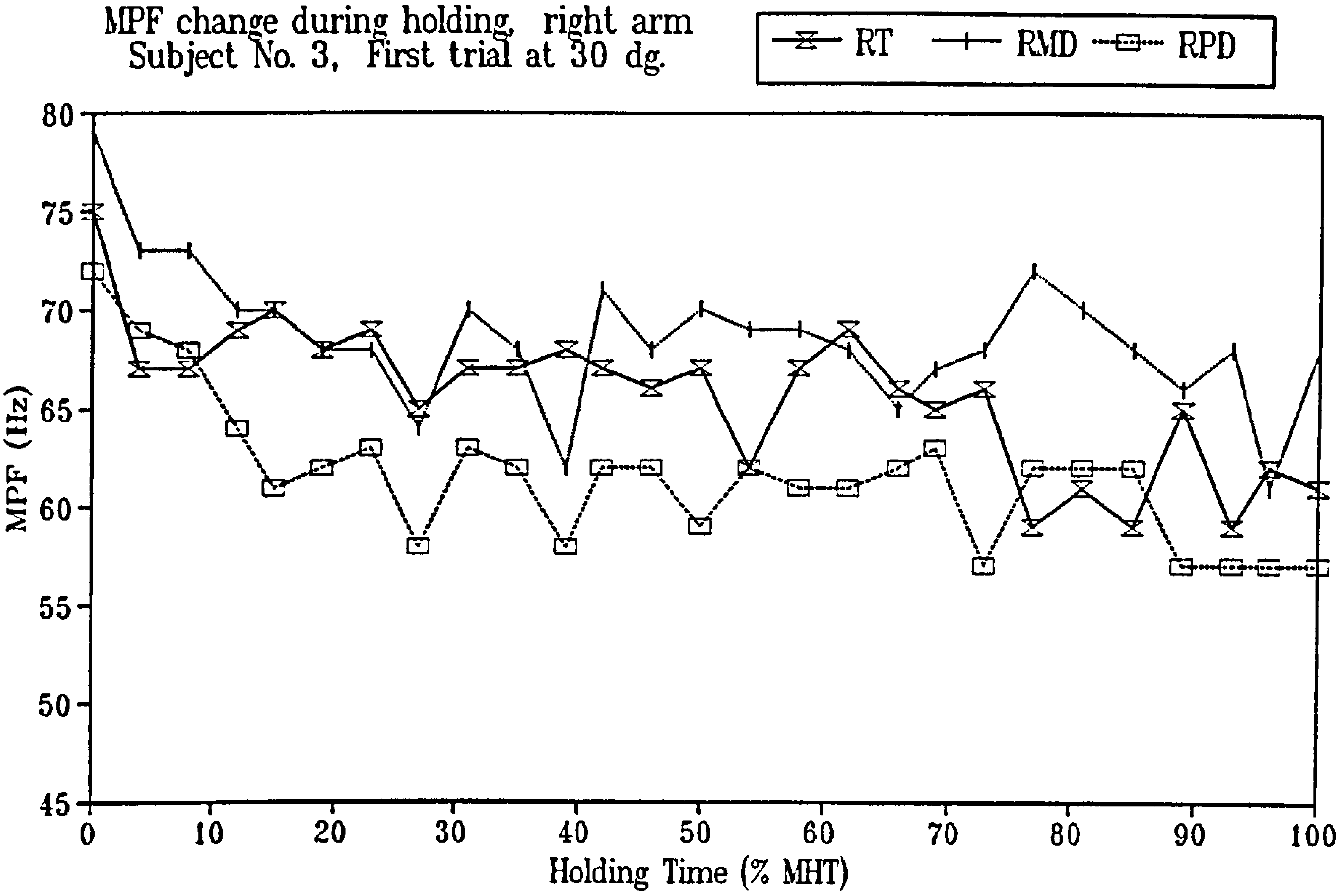


Fig D.32 Changes in MPF for subject No. 2 at 30 deg.



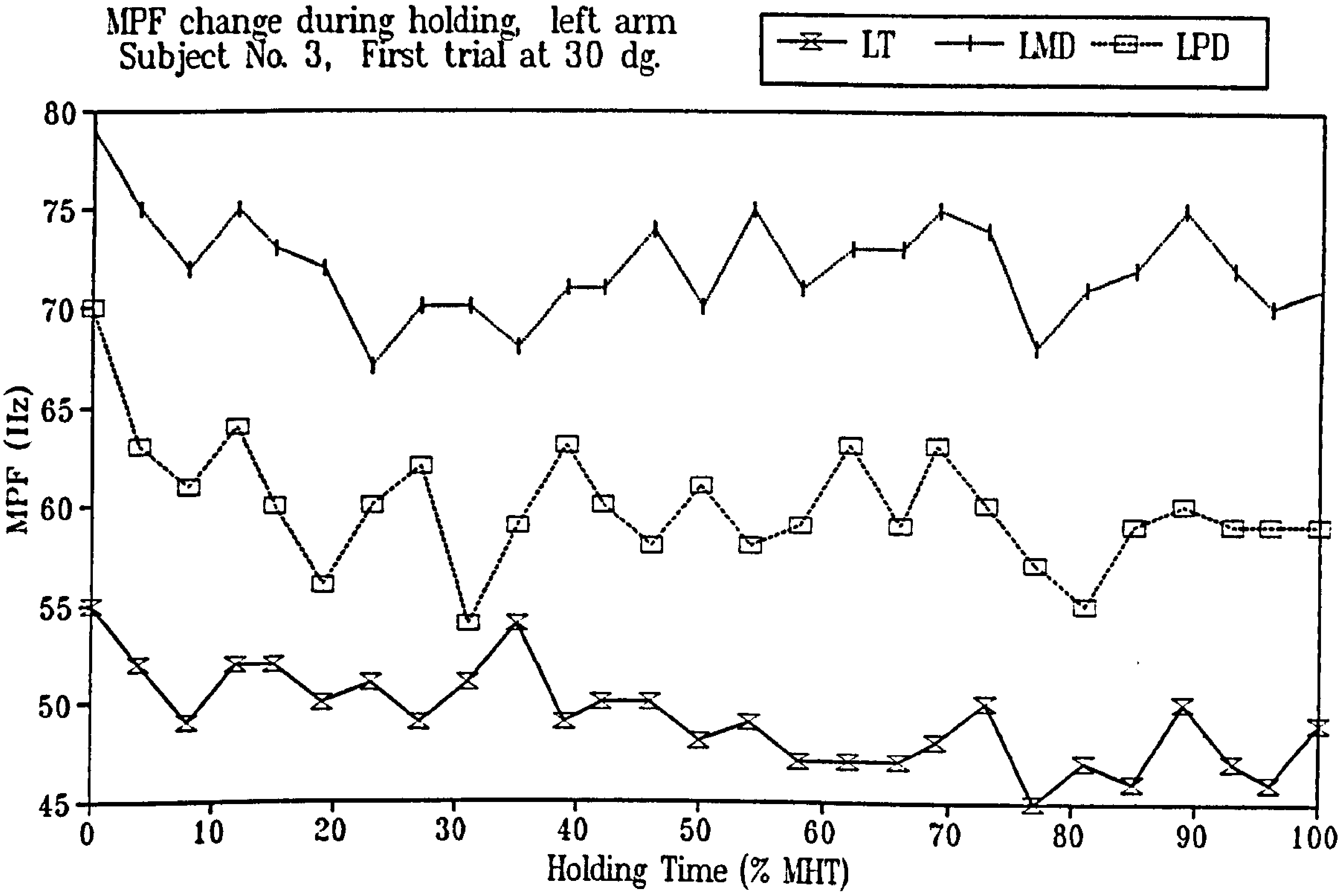
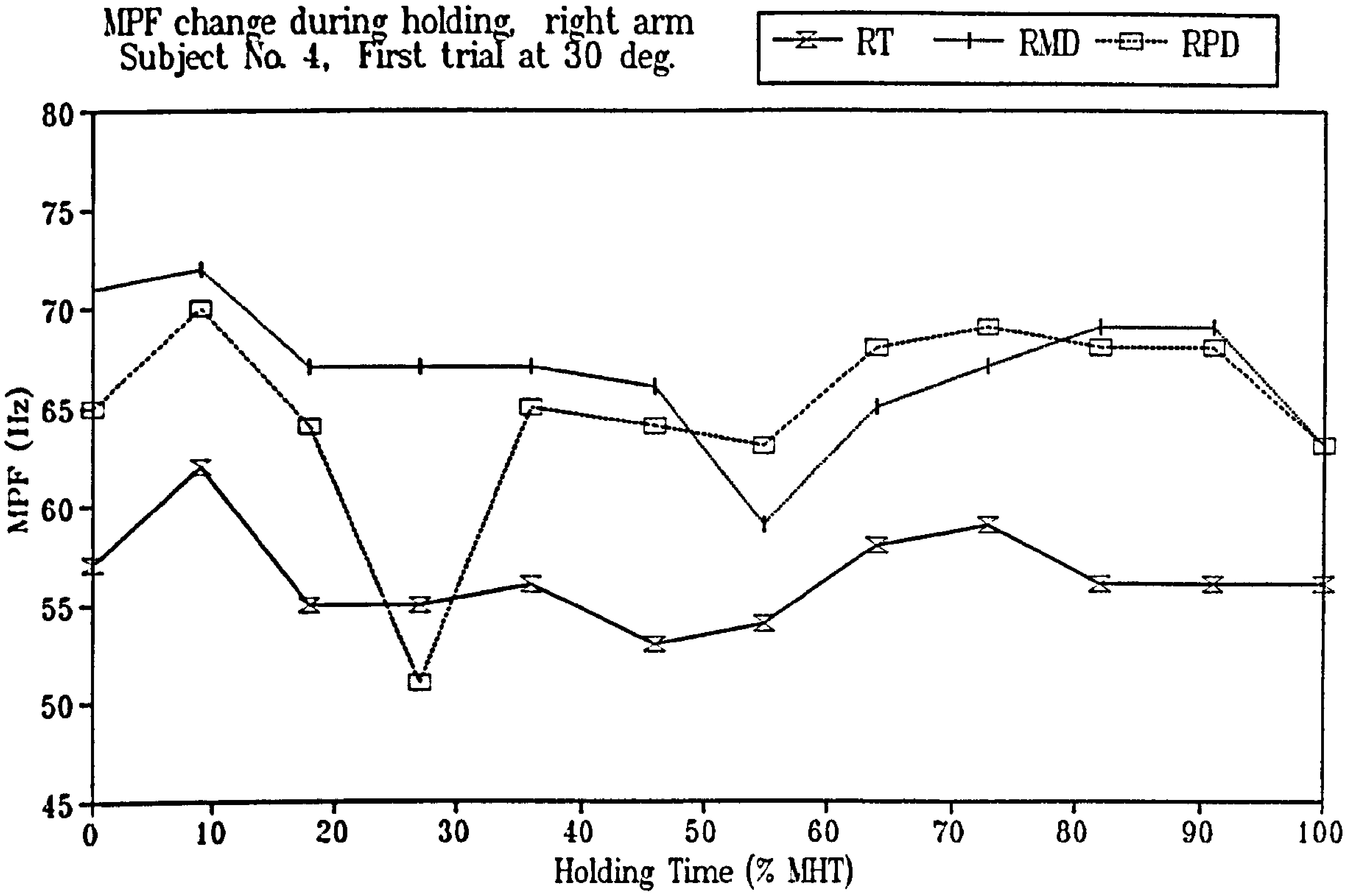


Figure D.33 Changes in MPF for subject No. 3 at 30 dg.



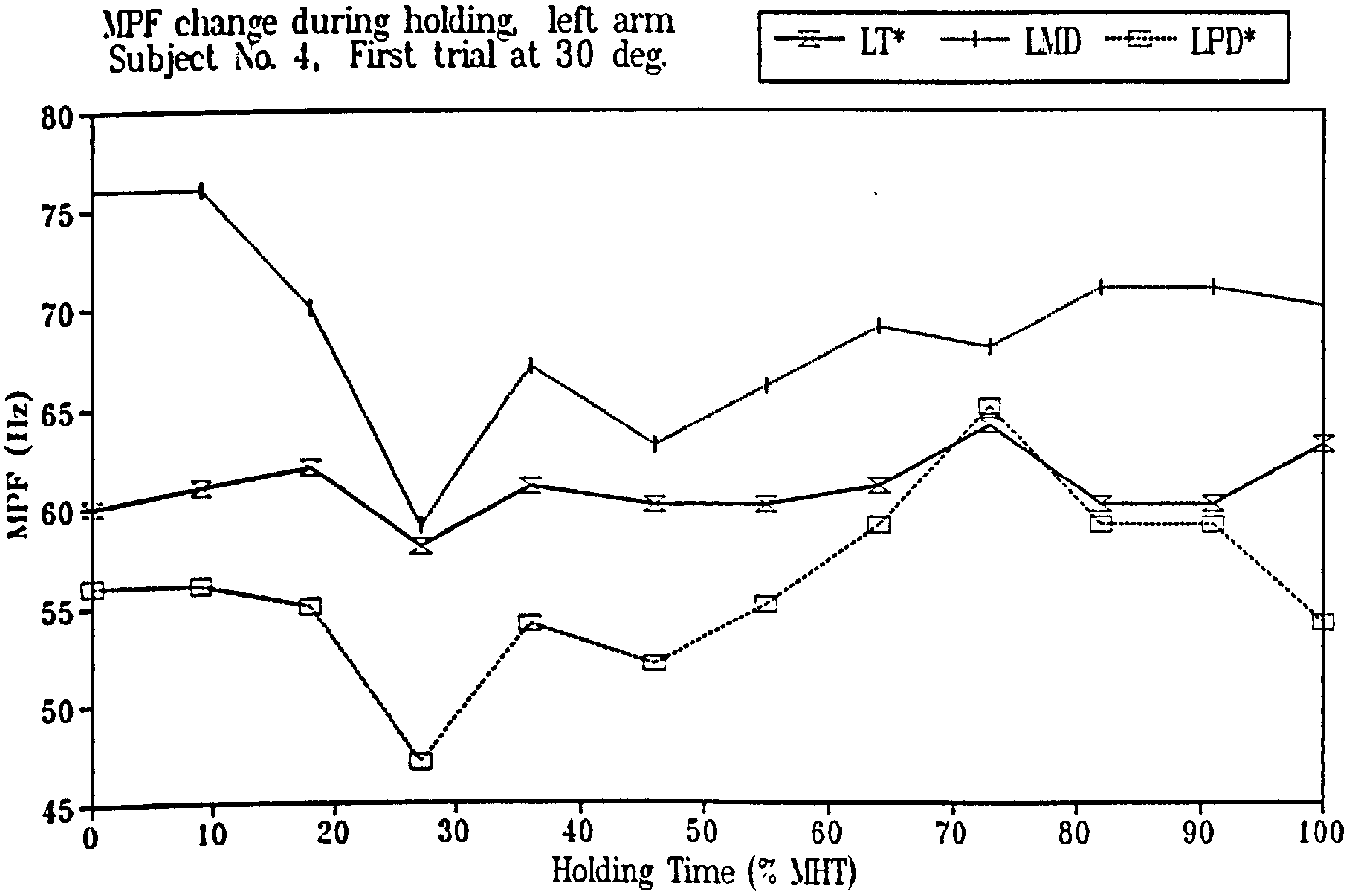
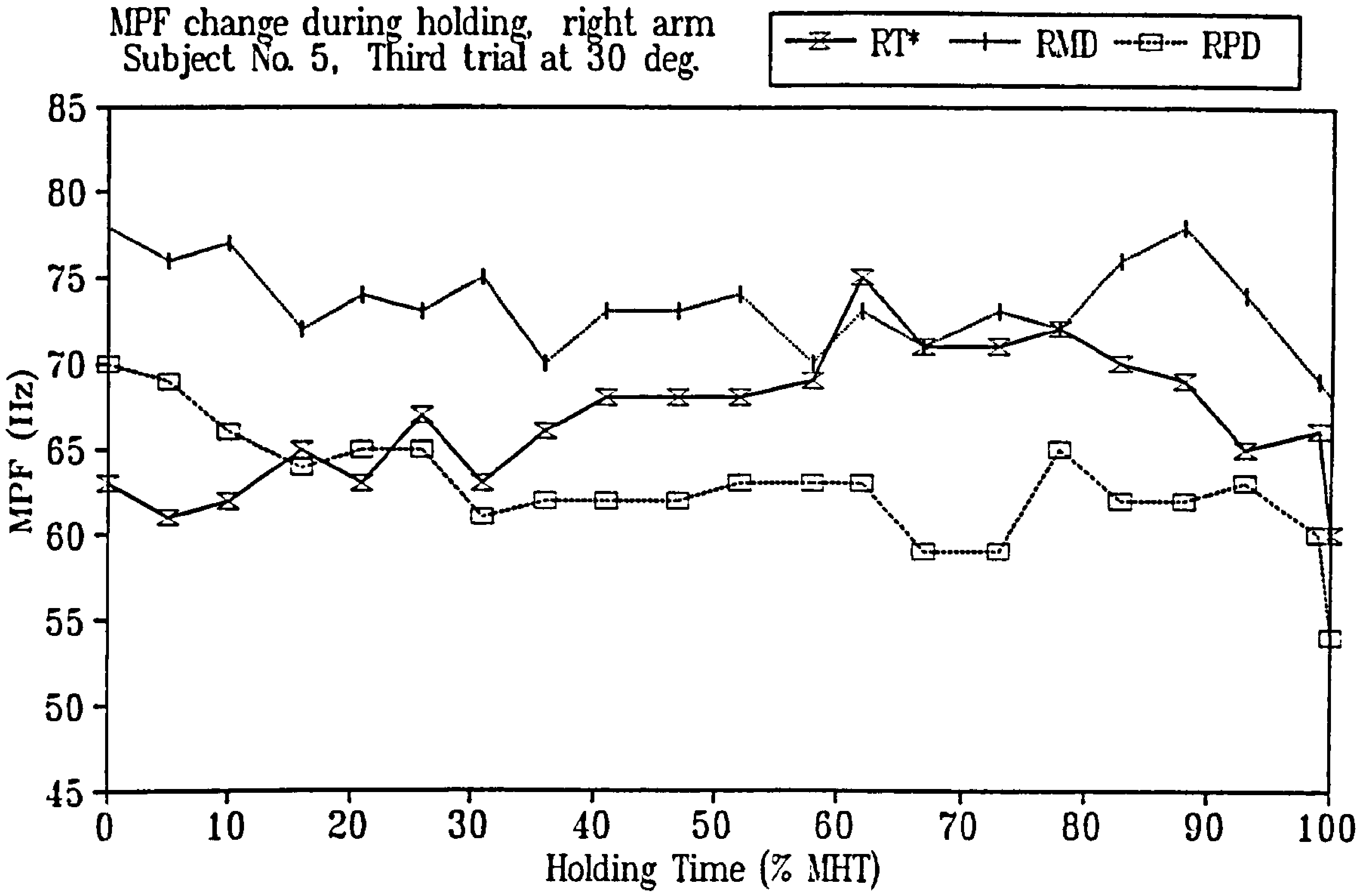


Figure D.34 Changes in MPF for subject No 4 at 30 deg.



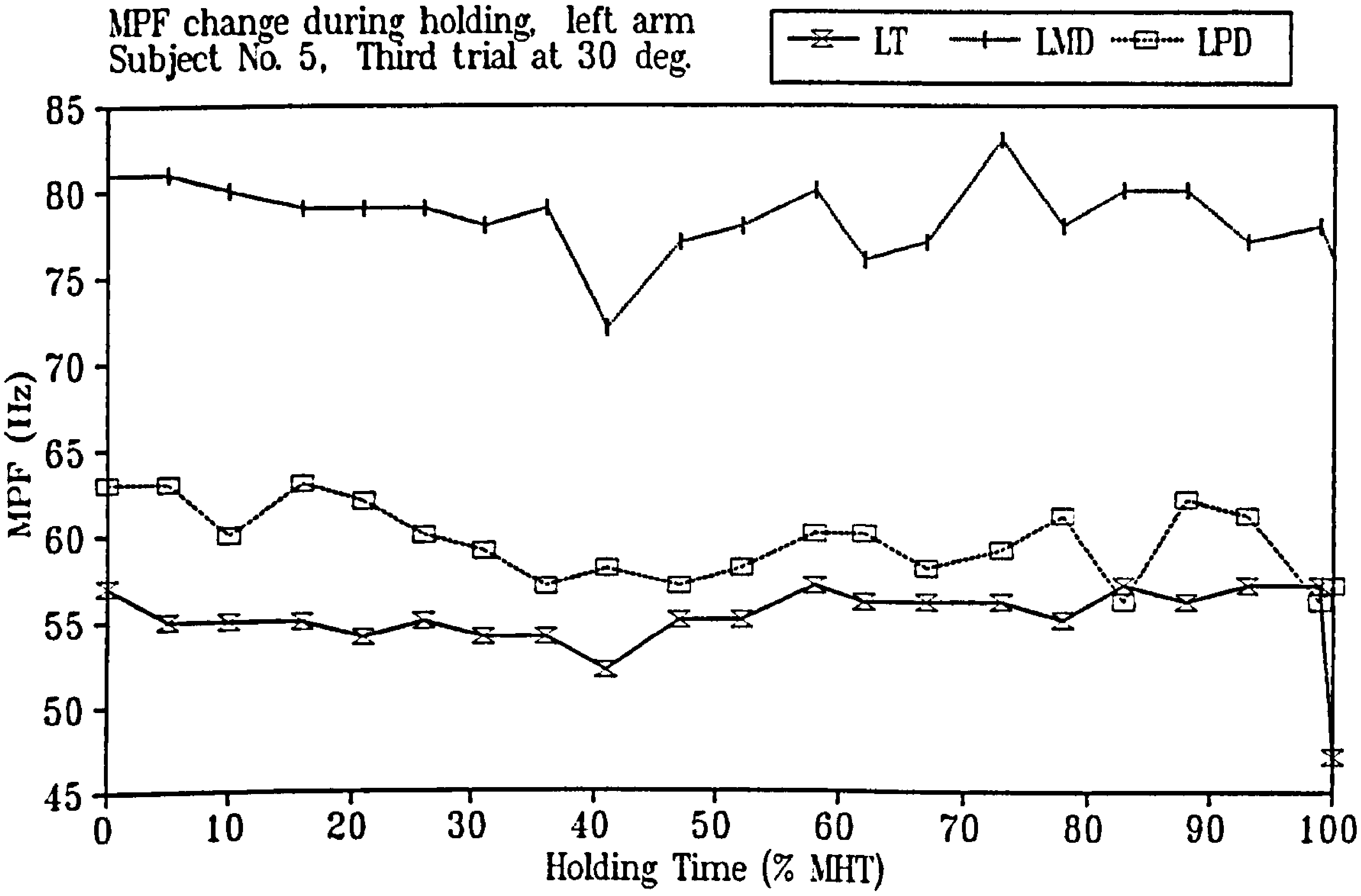


Fig D.35 Changes in MPF for subject No. 5 at 30 deg.

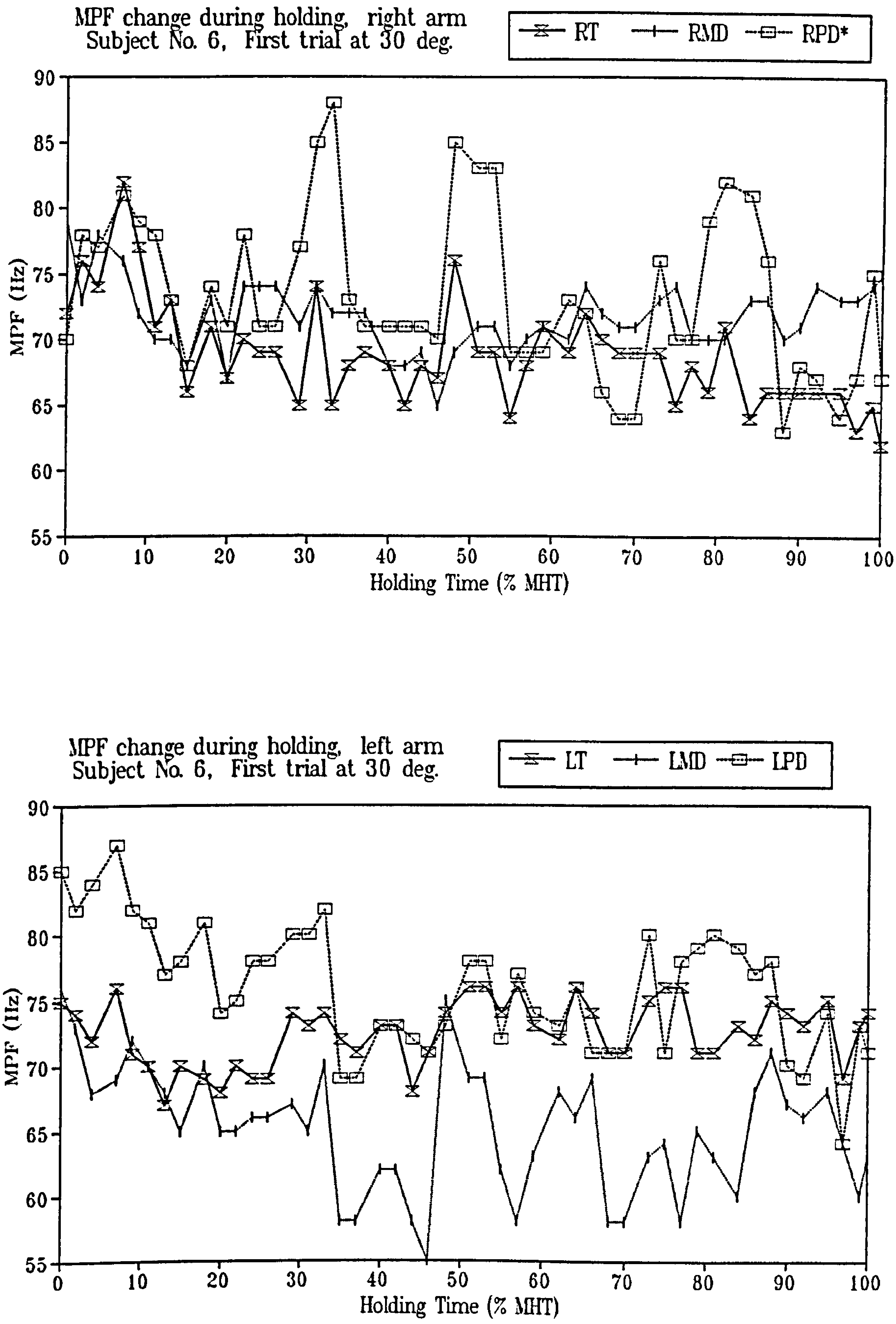
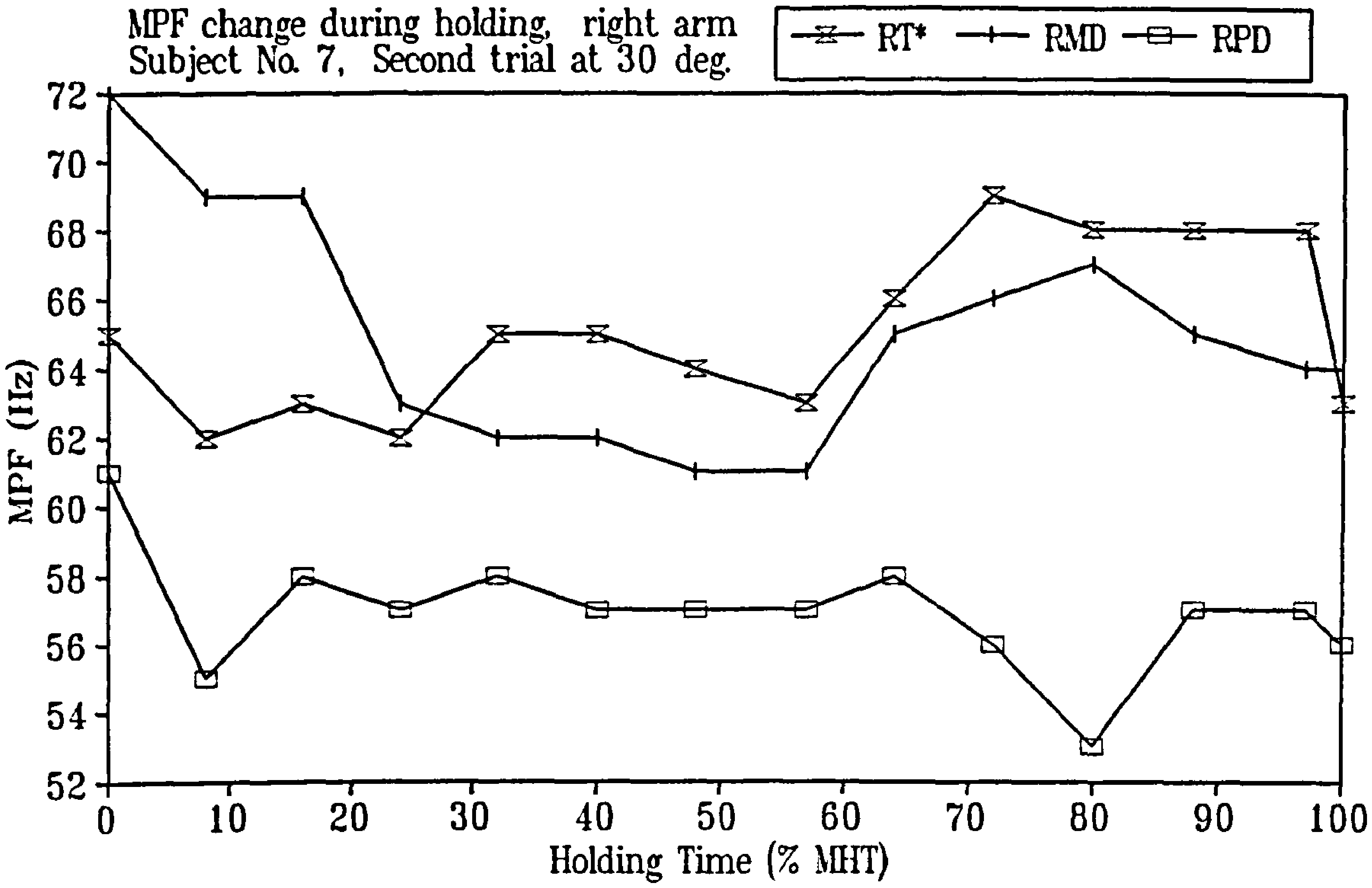


Fig D.36 Changes in MPF for subject No. 6 at 30 deg.



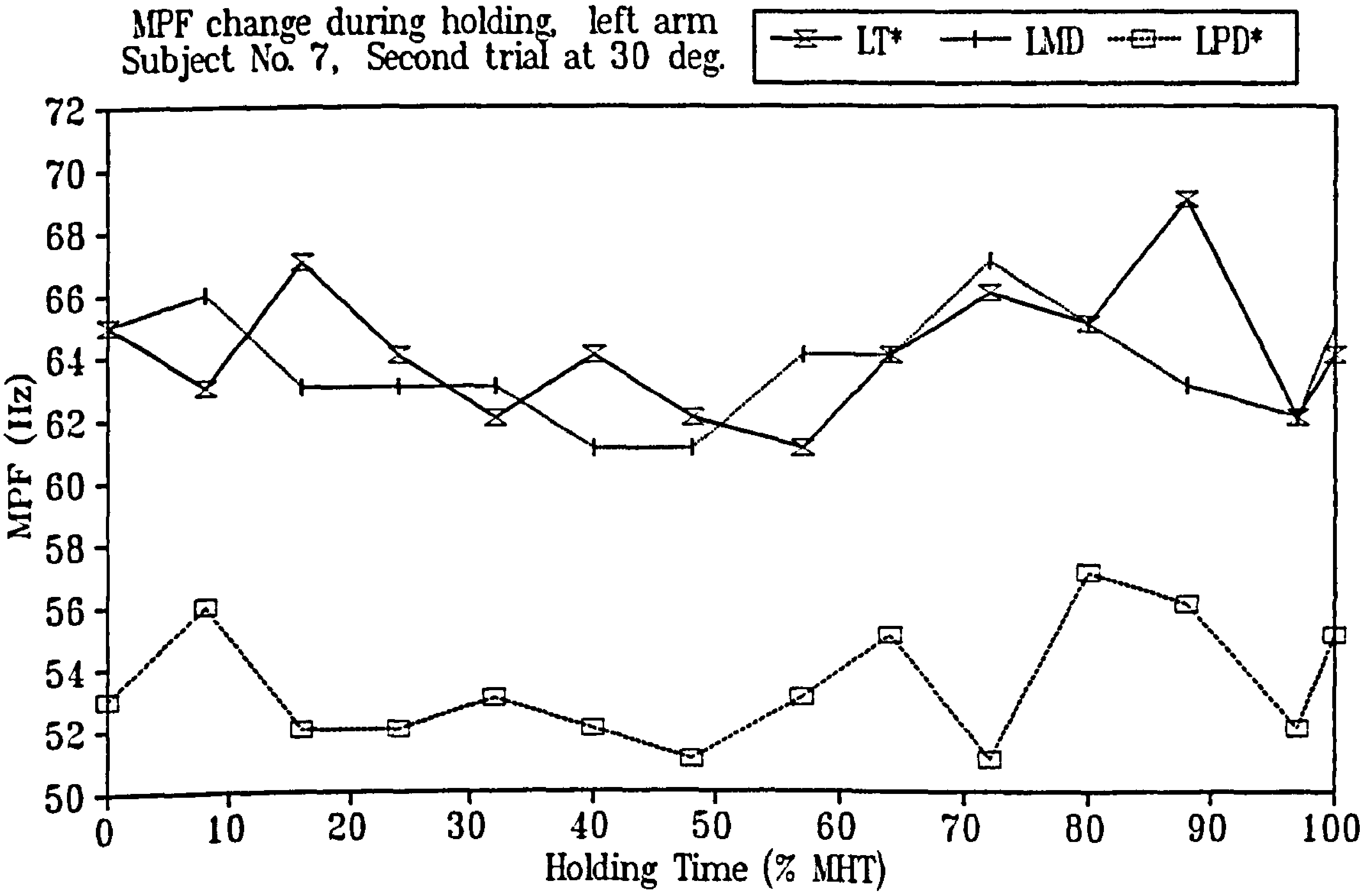
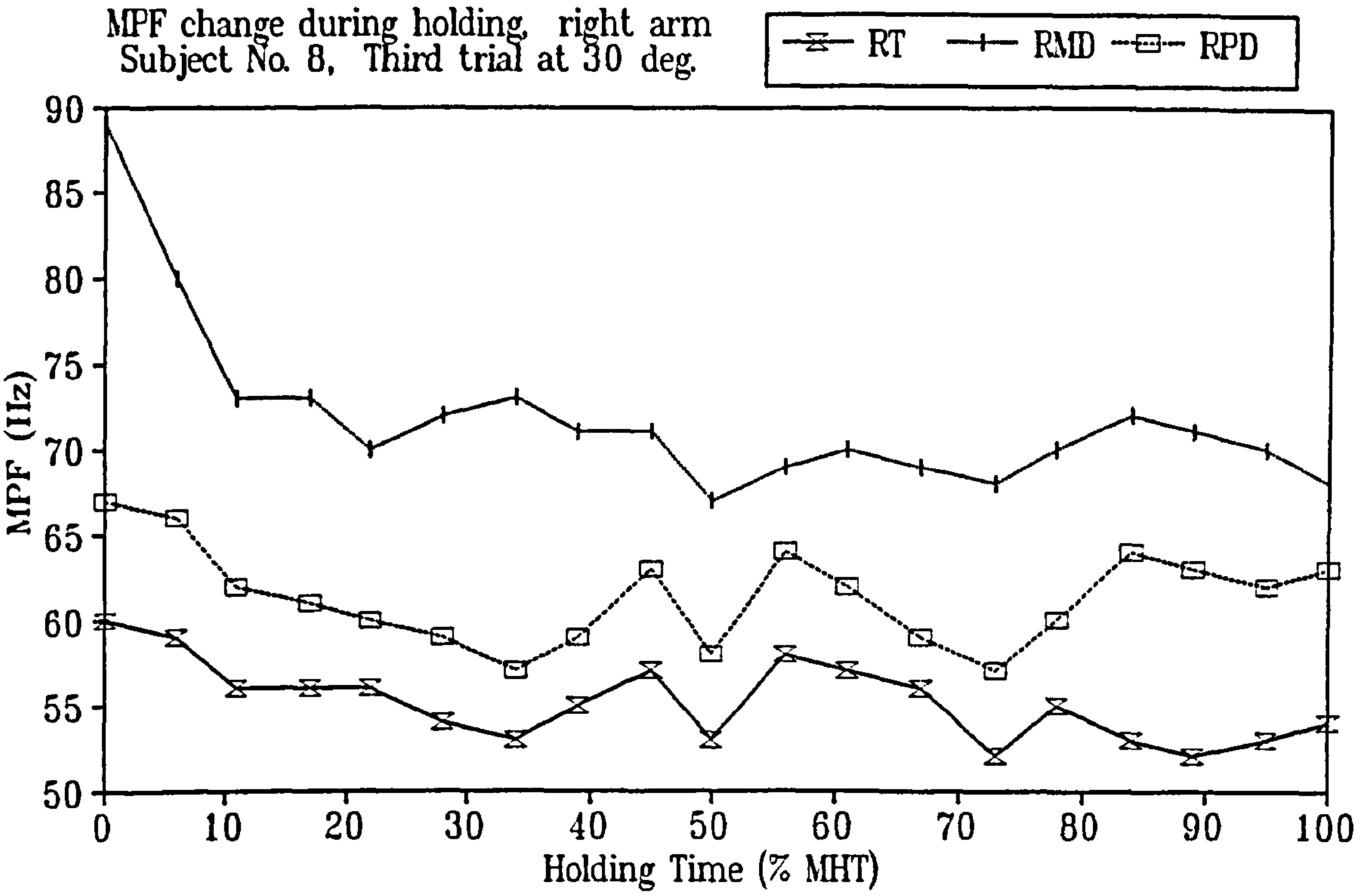


Figure D.37 Changes in MPF for subject No. 7 at 30 deg..



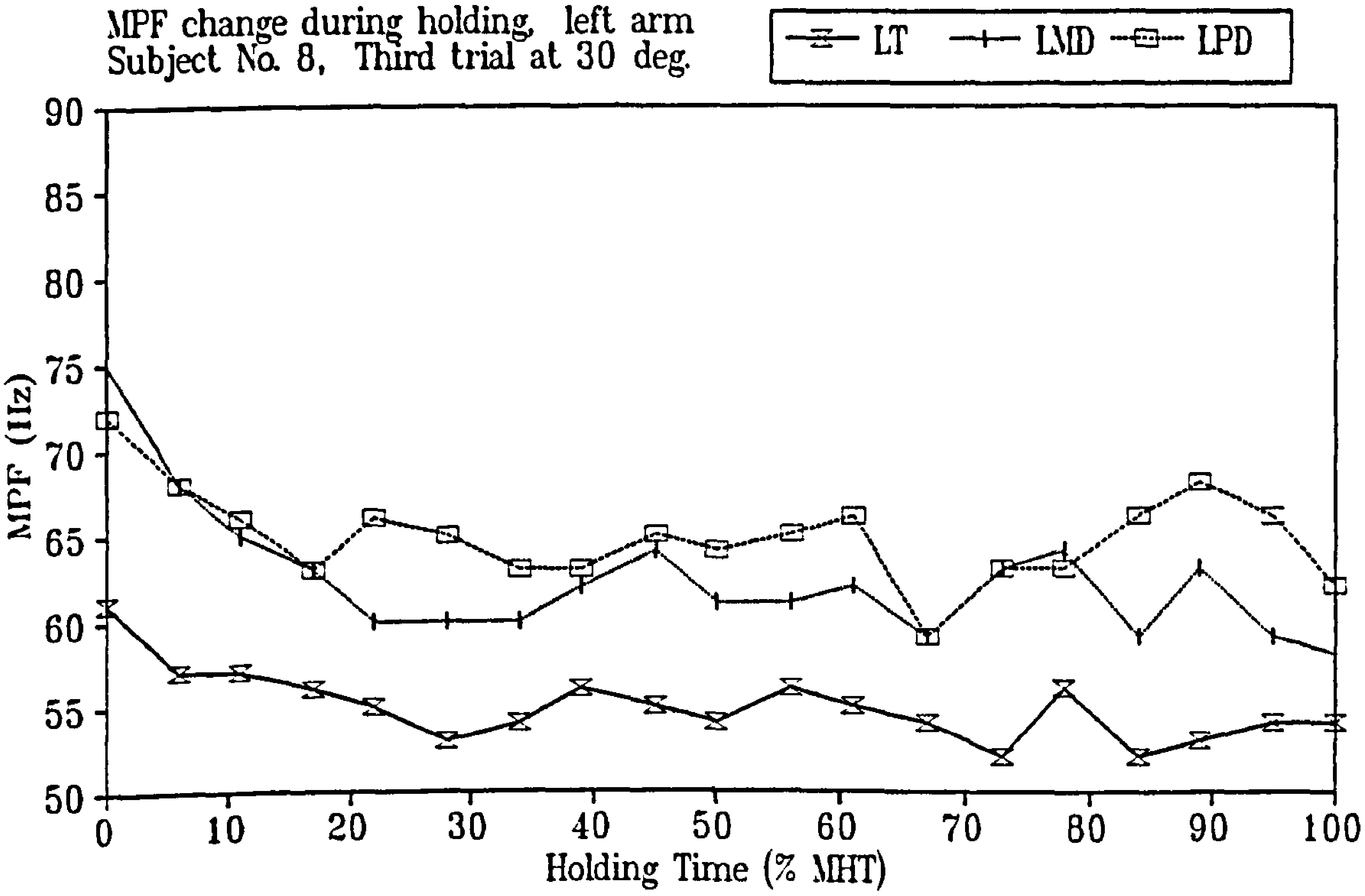
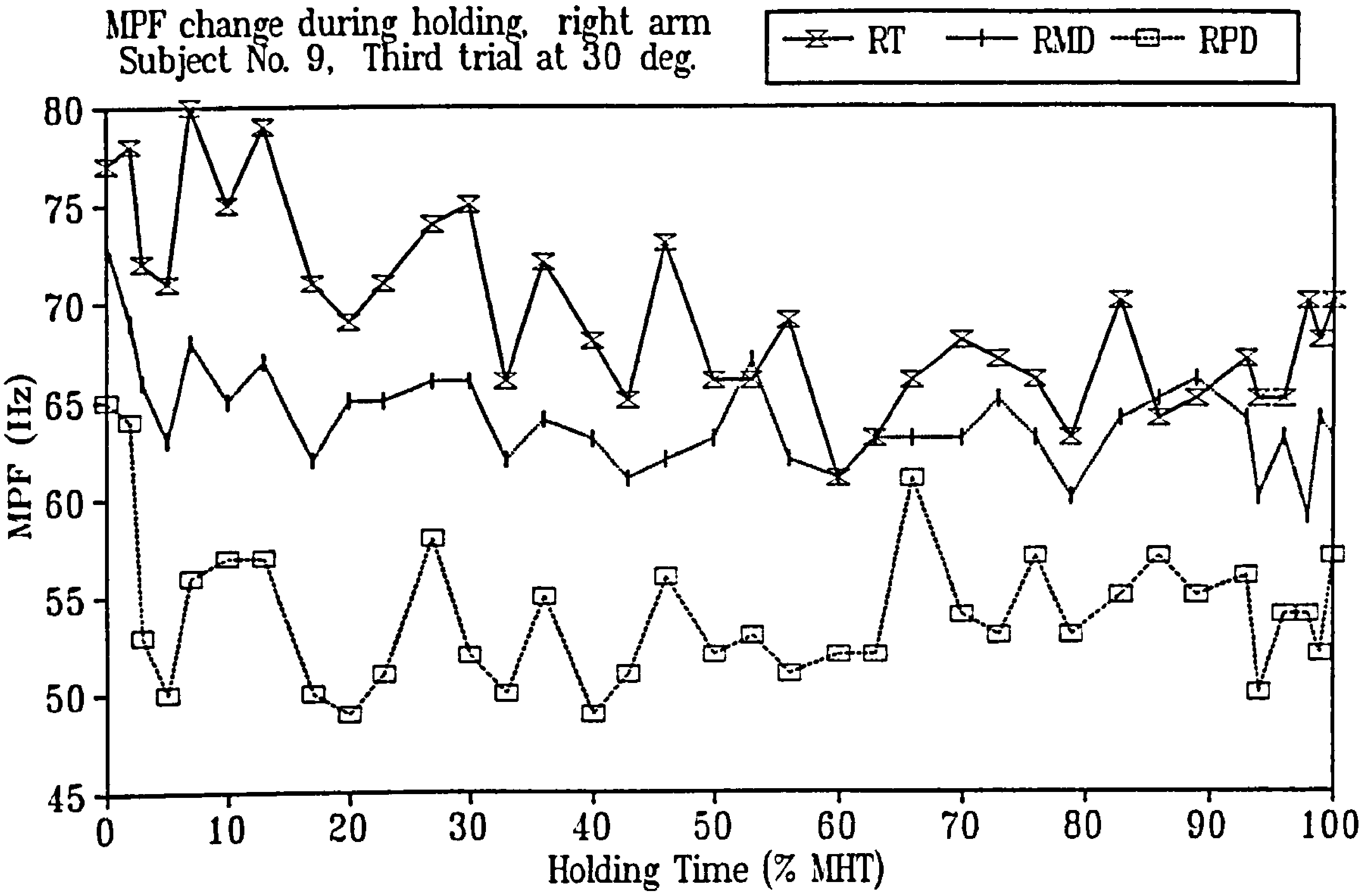


Figure D.37 Changes in MPF for subject No. 8 at 30 deg.



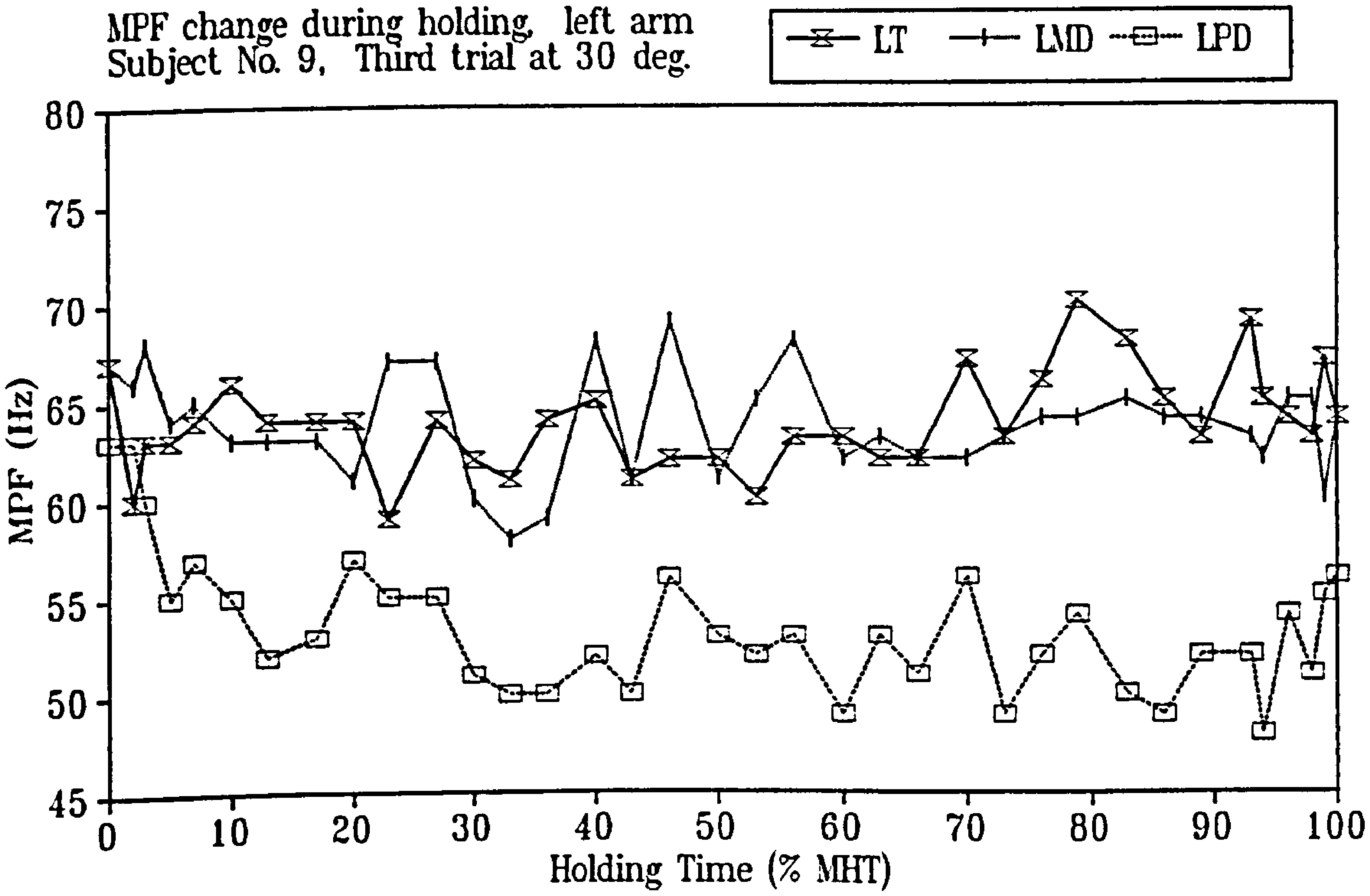
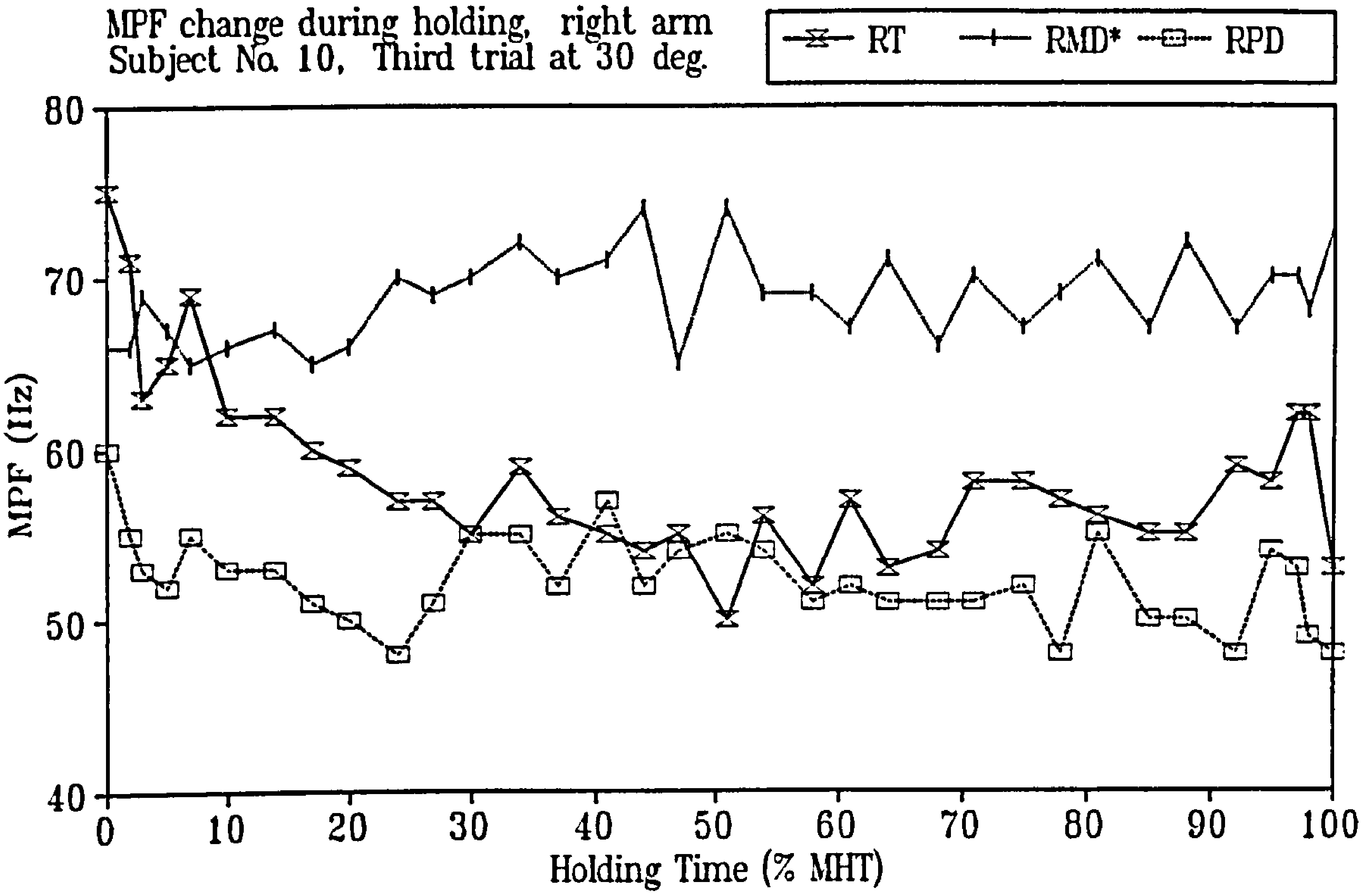


Figure D.39 Changes in MPF for subject No. 9 at 30 deg.


MPF change during holding, right arm
Subject No. 10, Third trial at 30 deg.

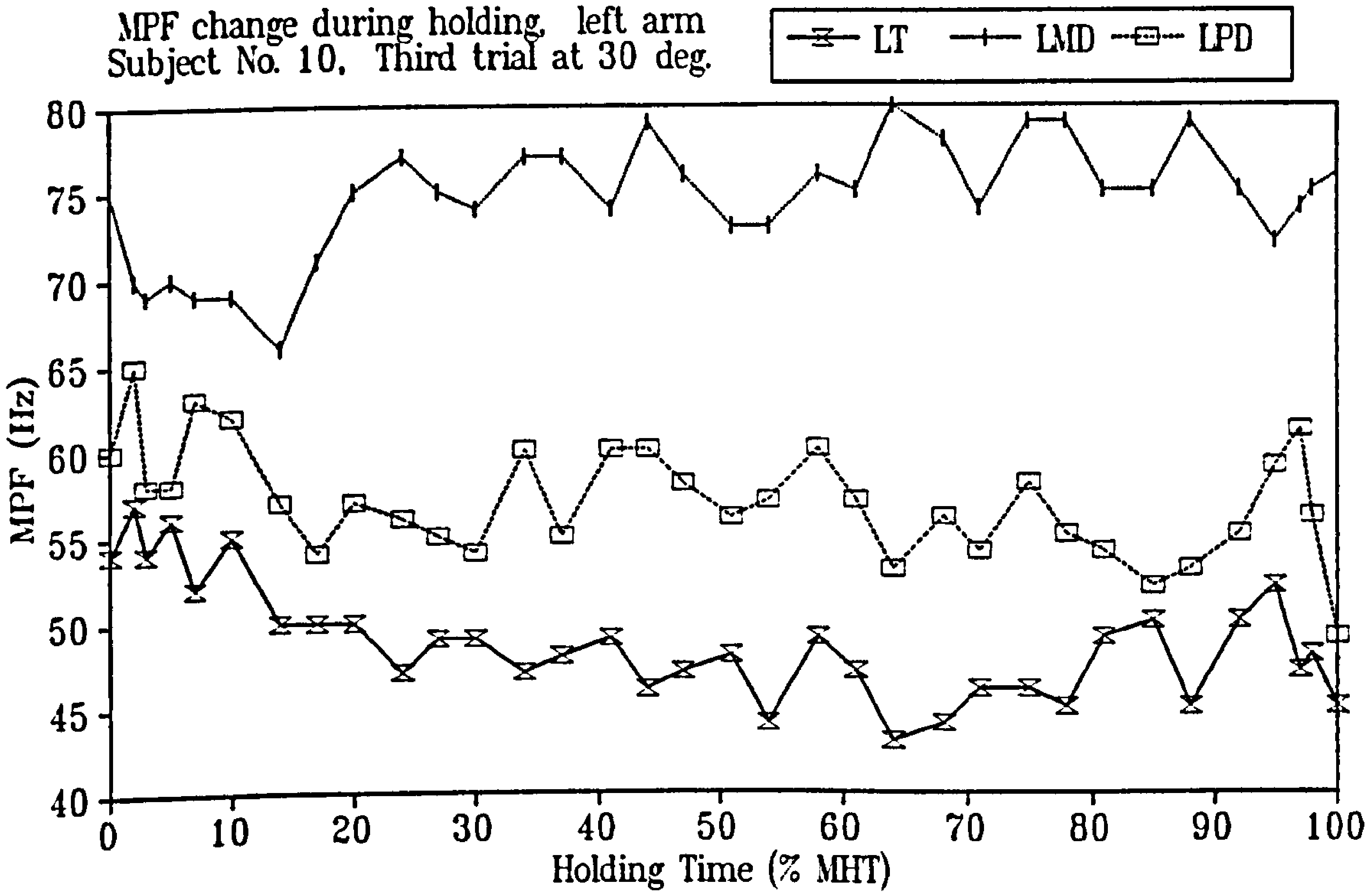
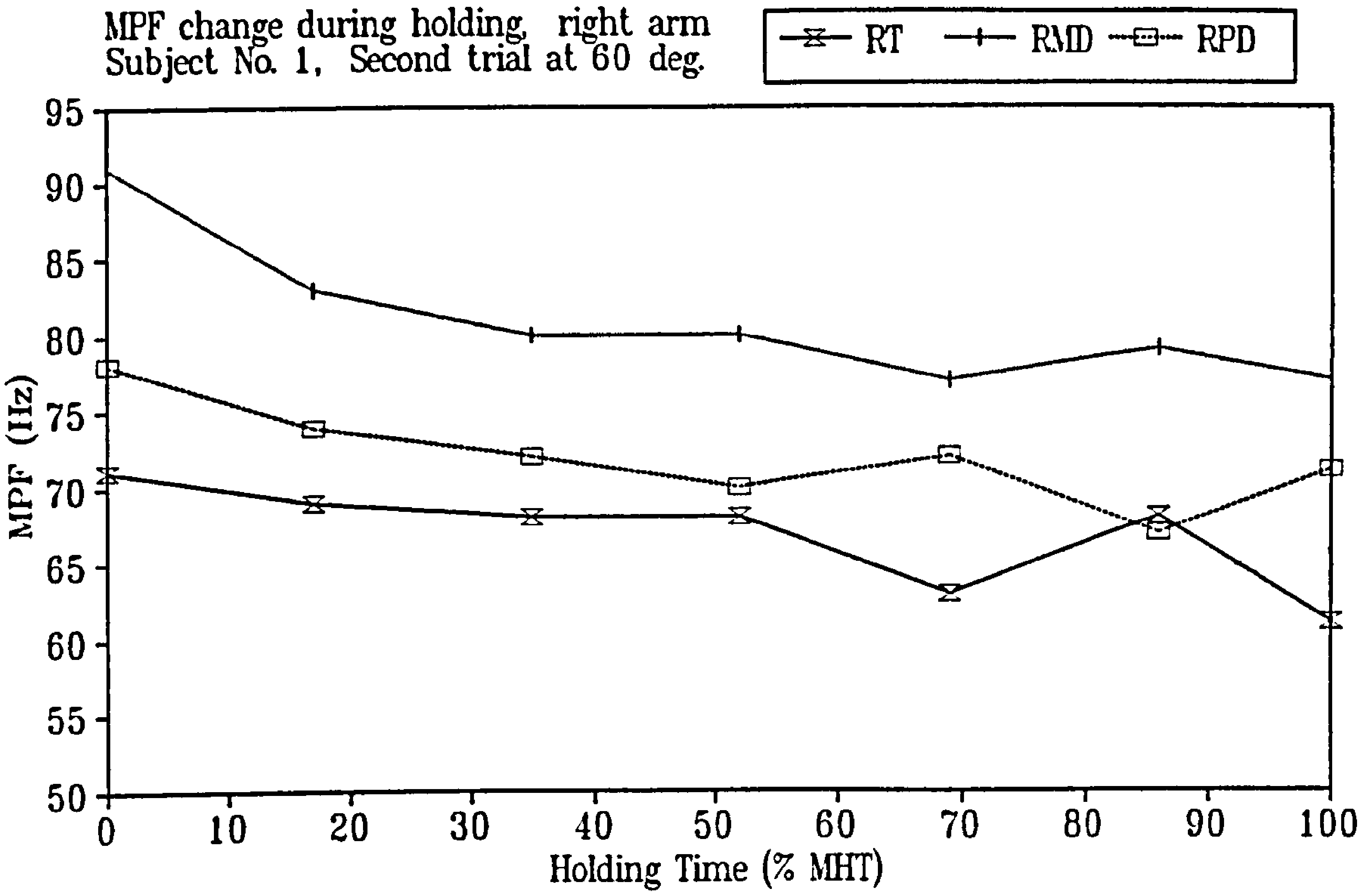

MPF change during holding, left arm
Subject No. 10, Third trial at 30 deg.

Fig D.40 Changes in MPF for subject No. 10 at 30 deg.



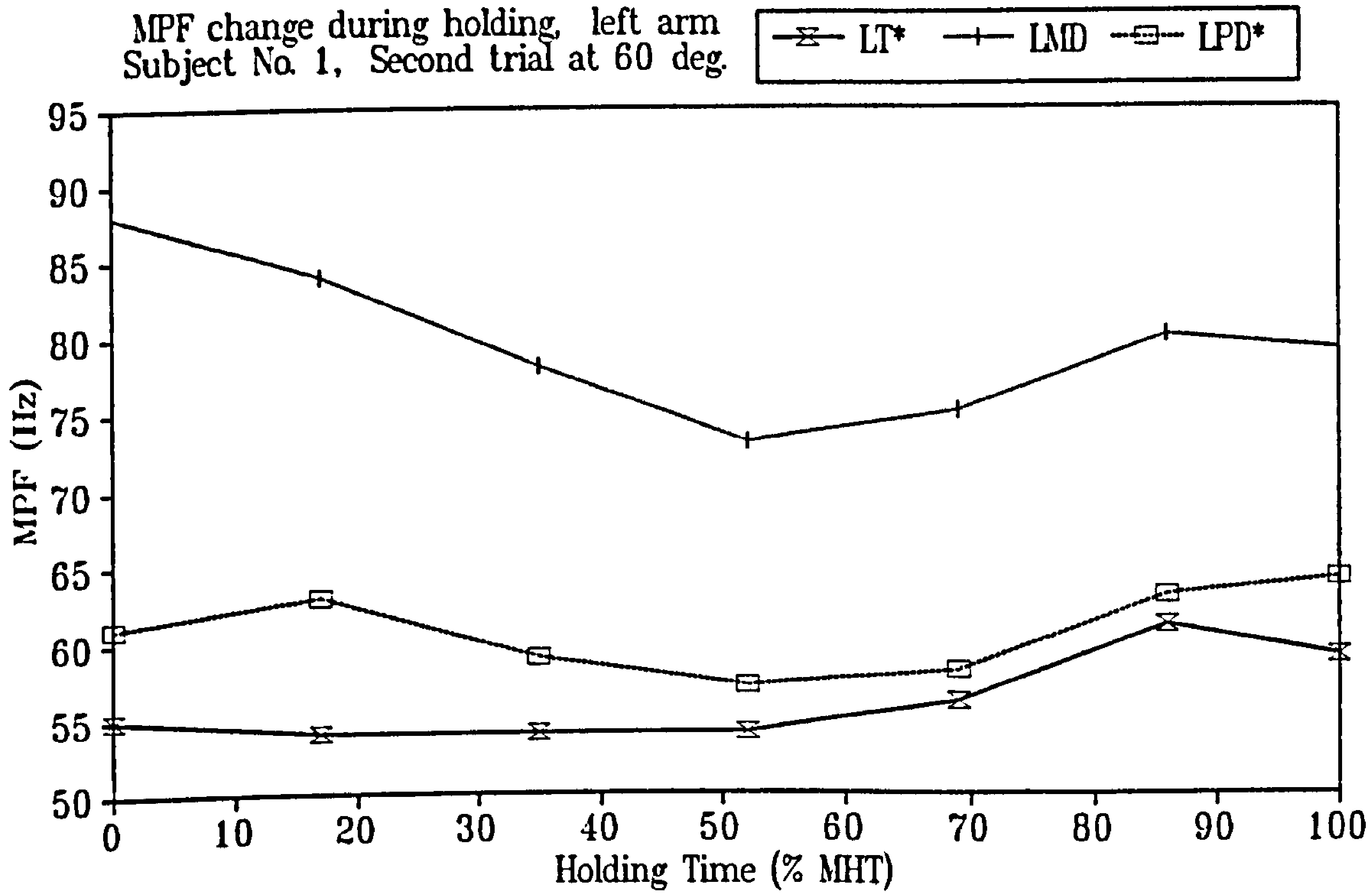
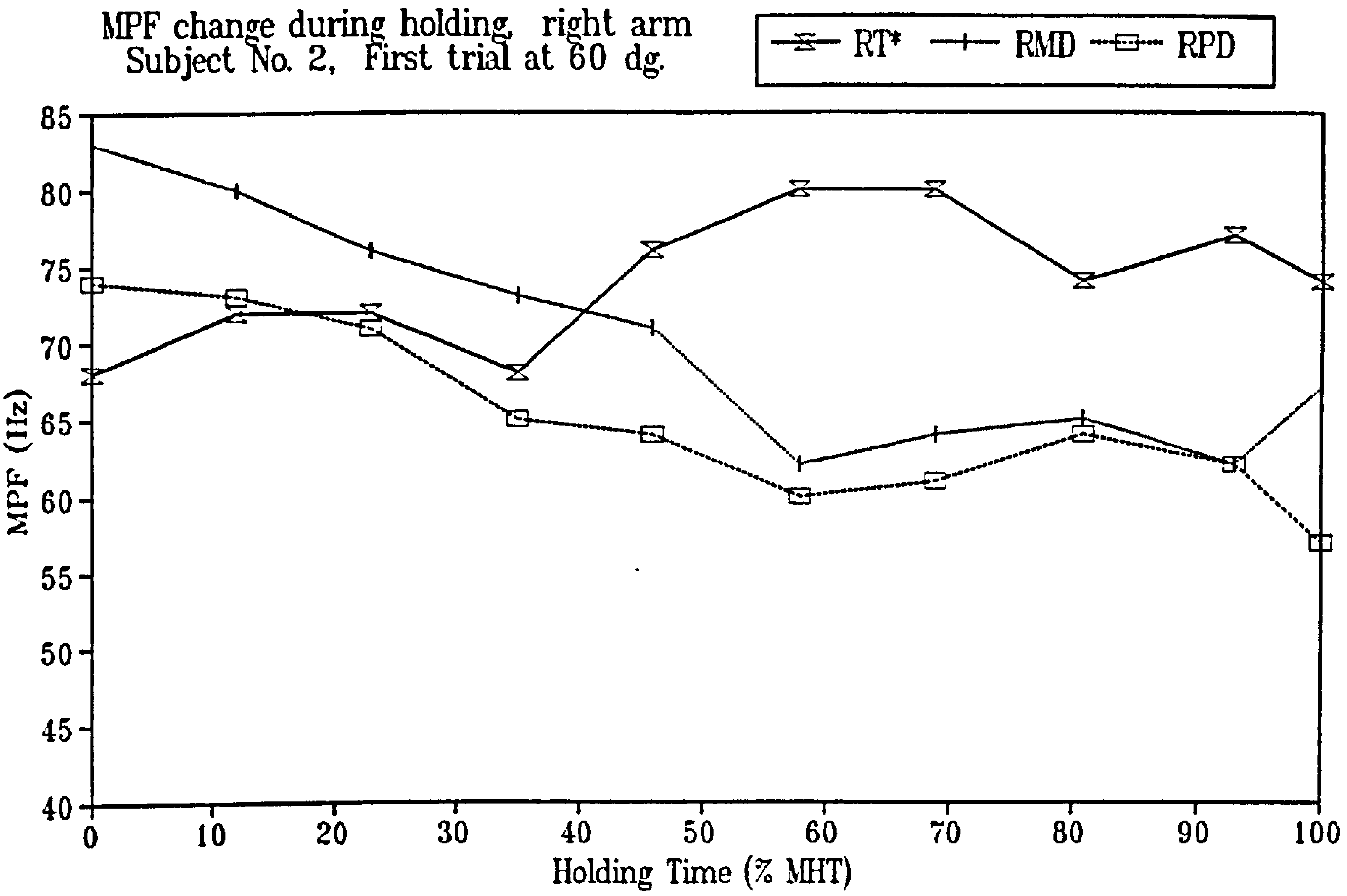


Figure D.41 Changes in MPF for subject No. 1 at 60 deg.



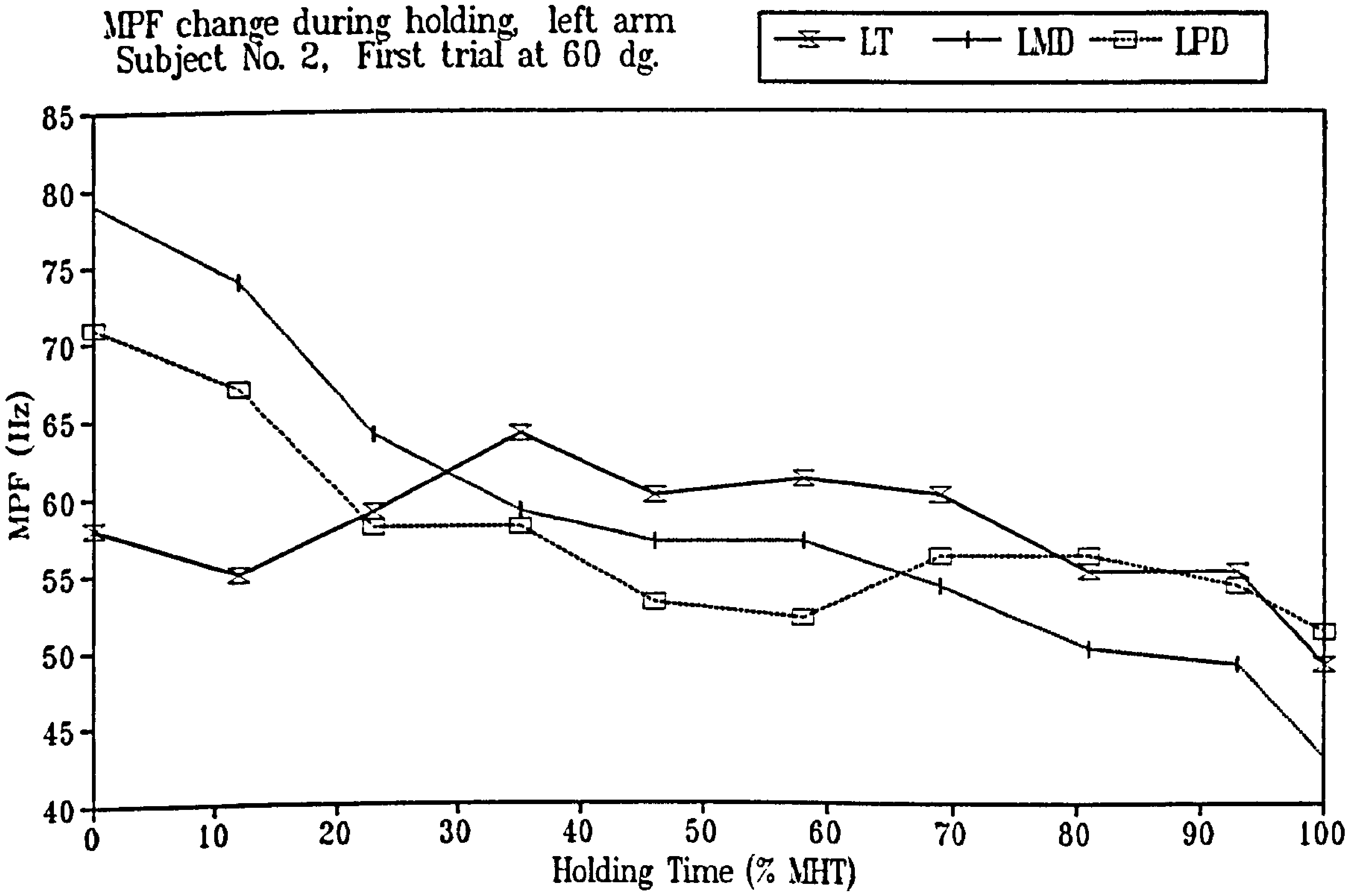
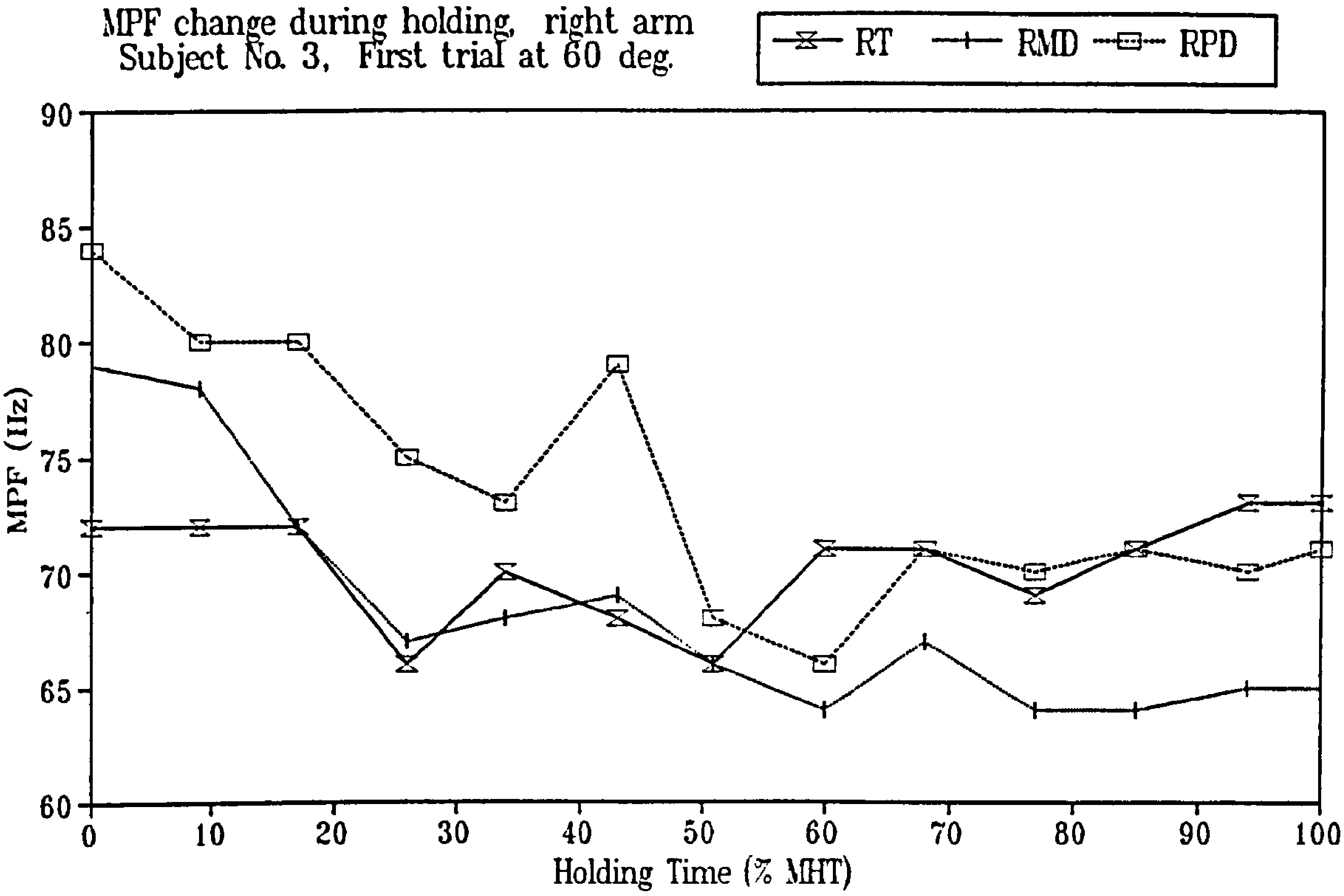


Figure D.42 Changes in MPF for subject No. 2 at 60 dg.



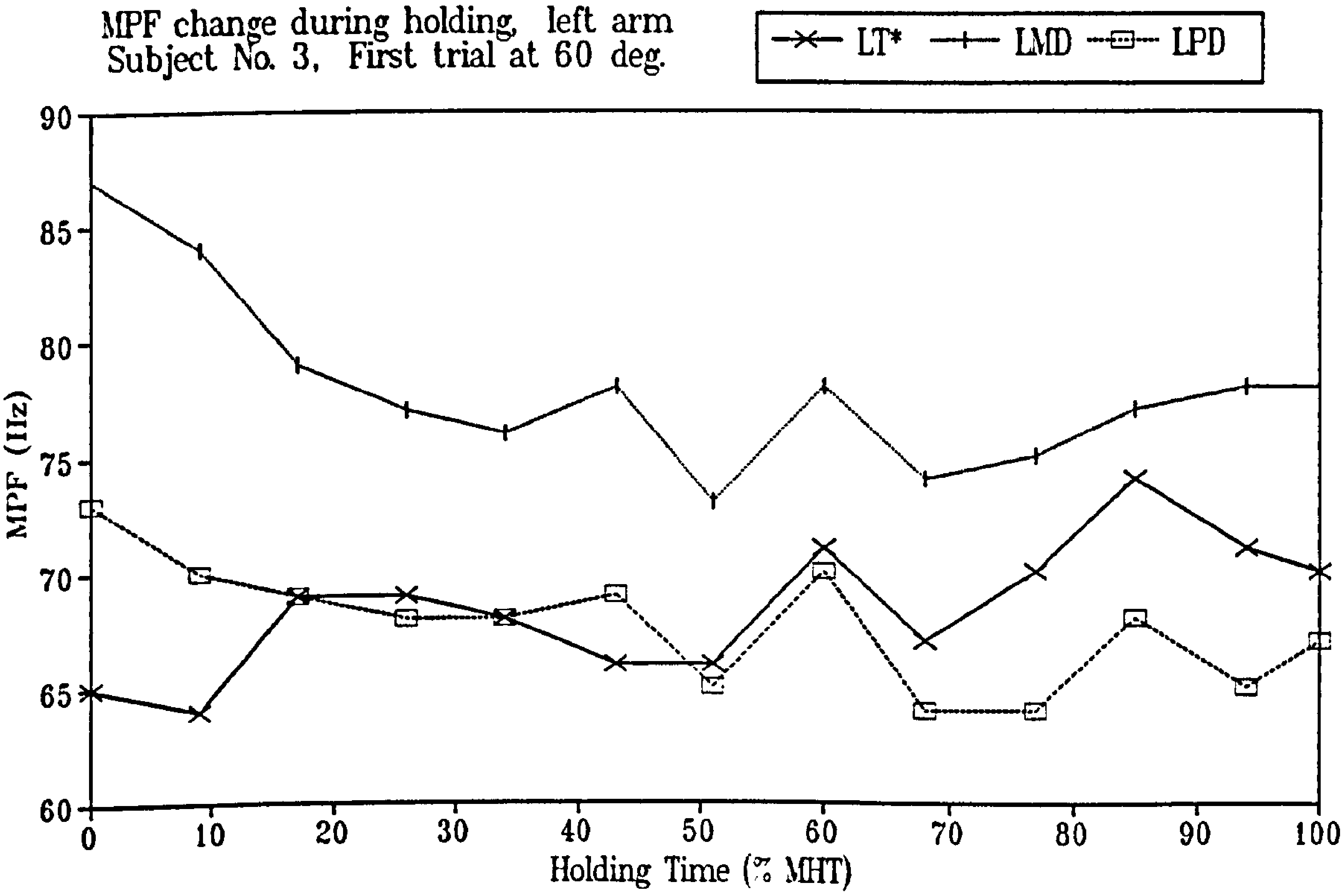
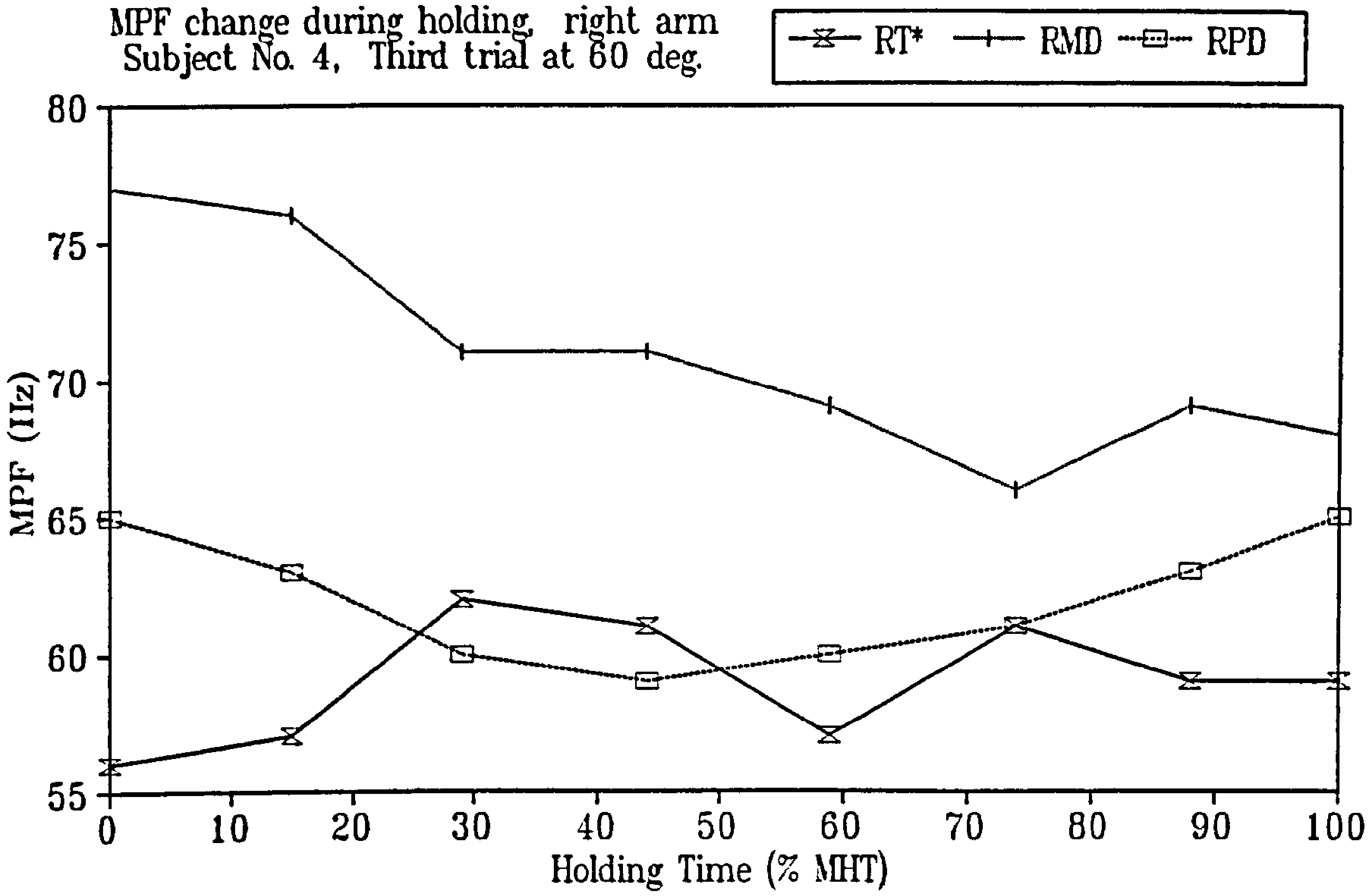


Figure D.43 Changes in MPF for subject No. 3 at 60 dg.



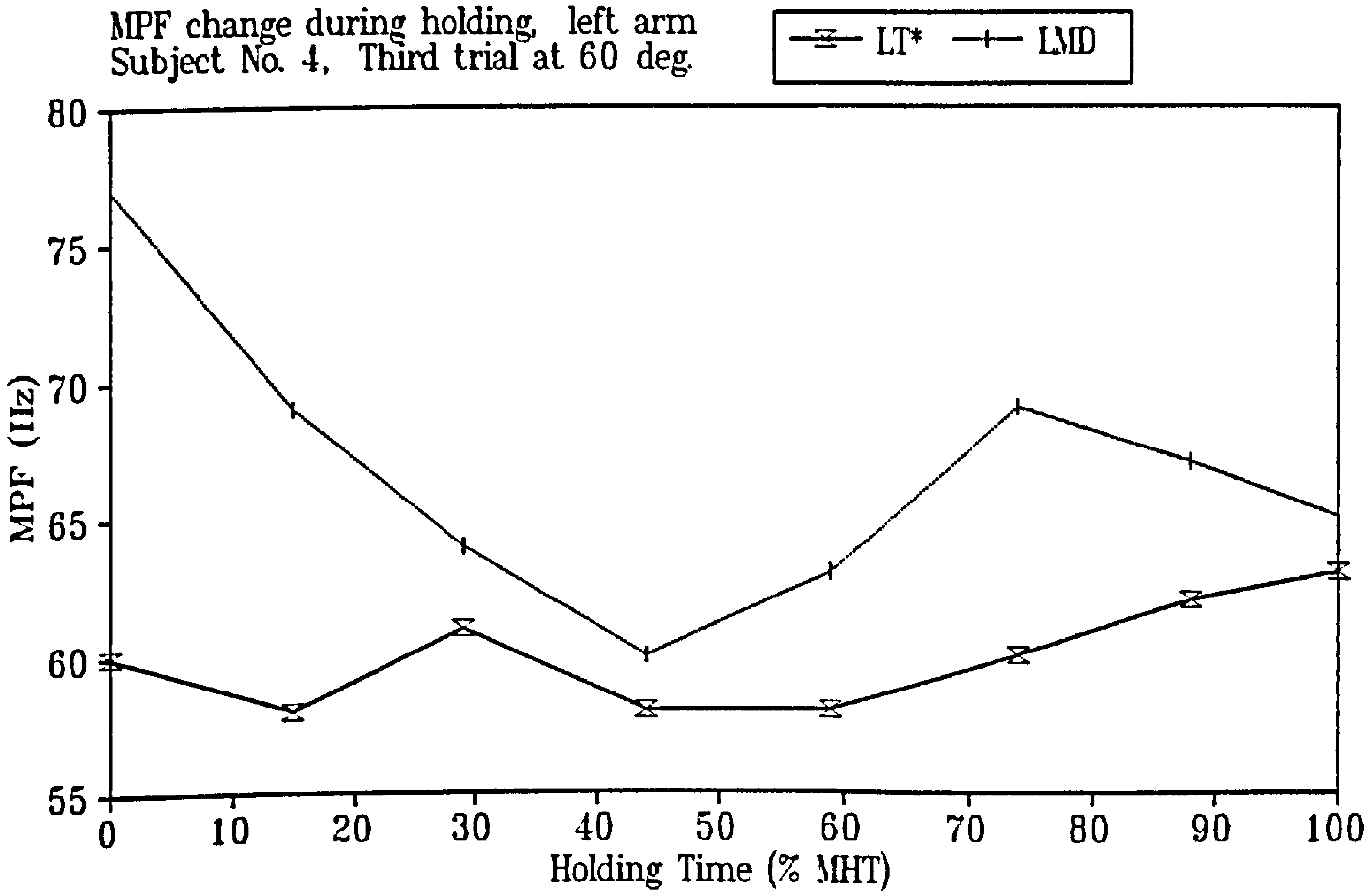


Fig D.44 Changes in MPF for subject No. 4 at 60 deg.

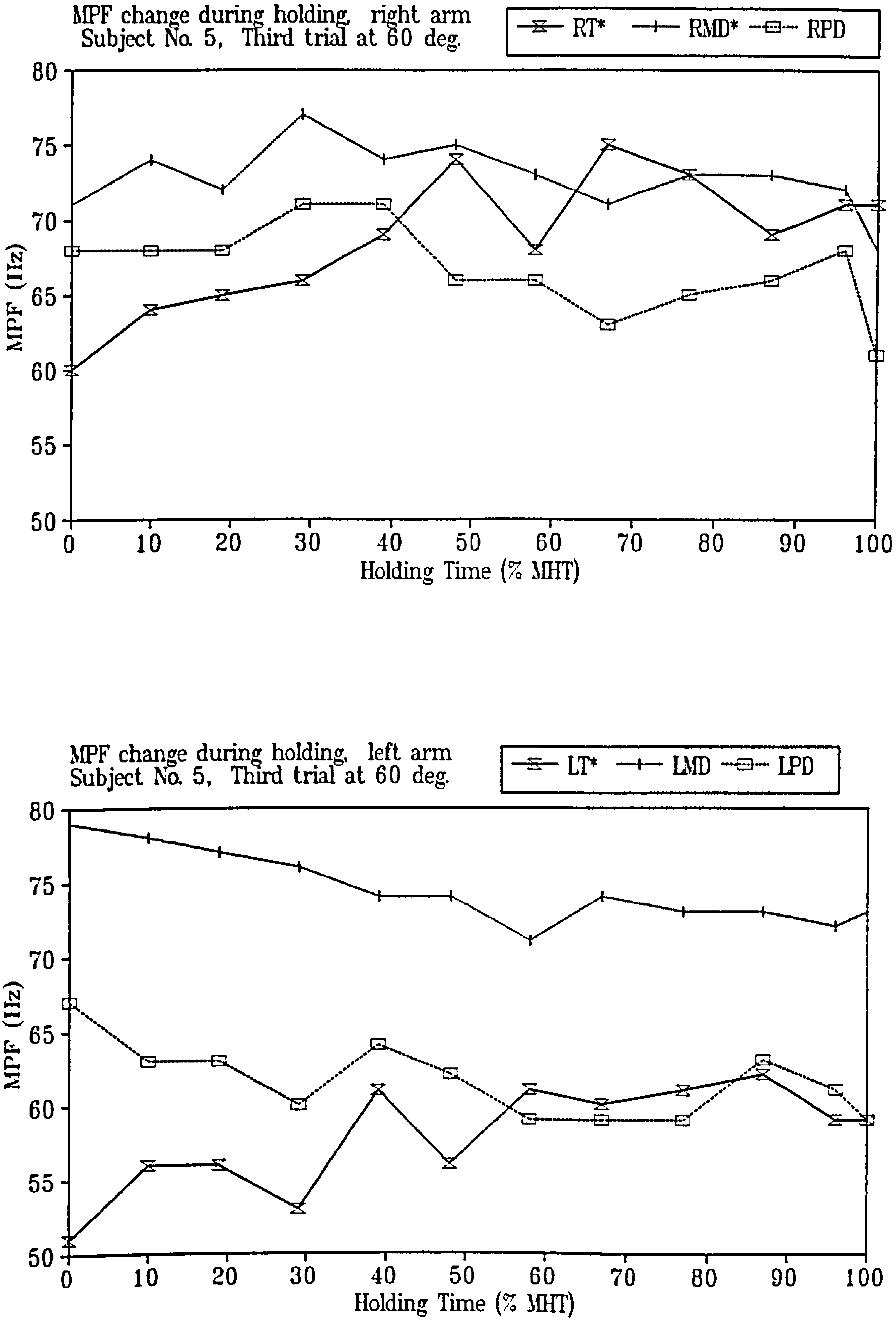


Figure D.55 Changes in MPF for subject No. 5 at 60 deg.

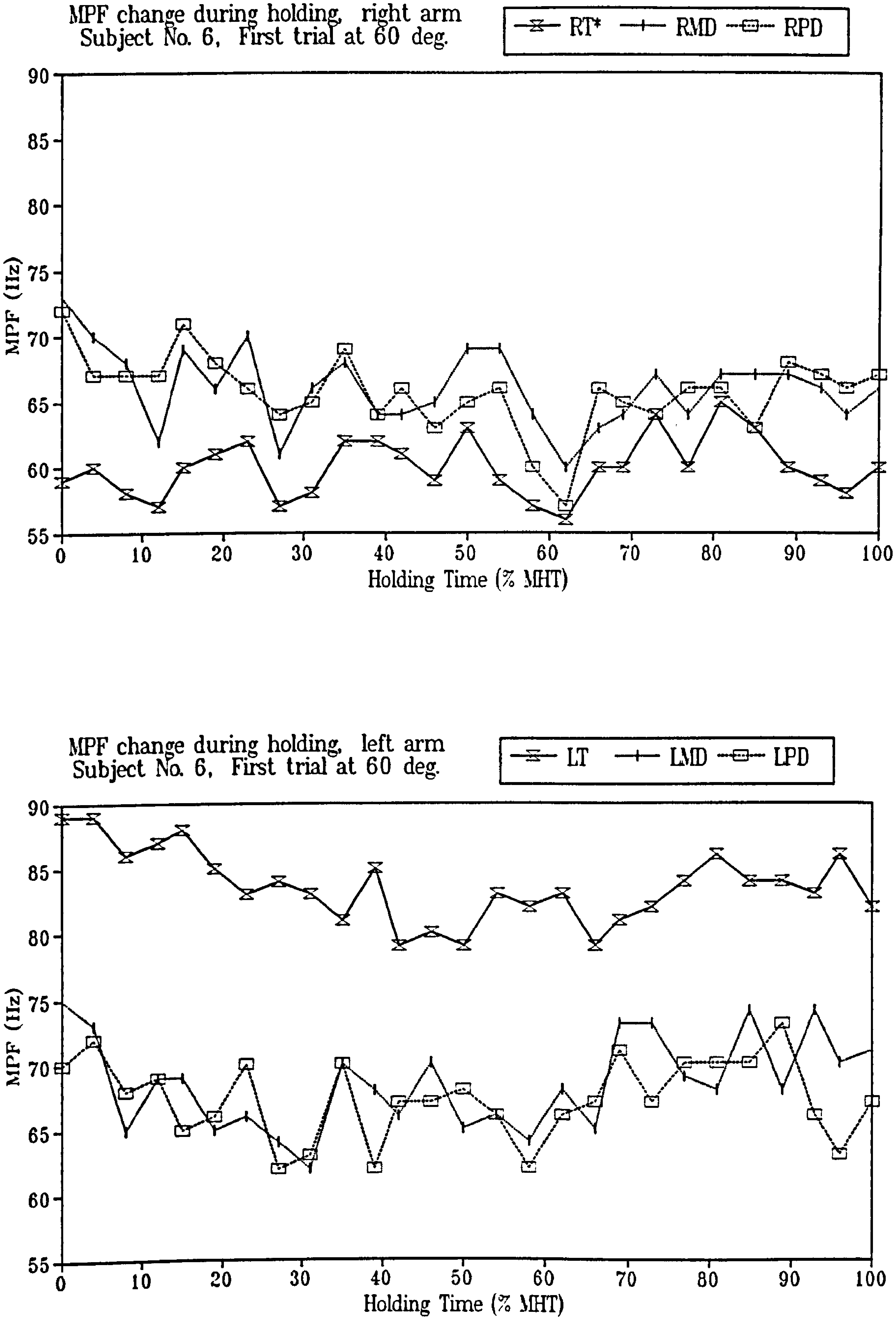
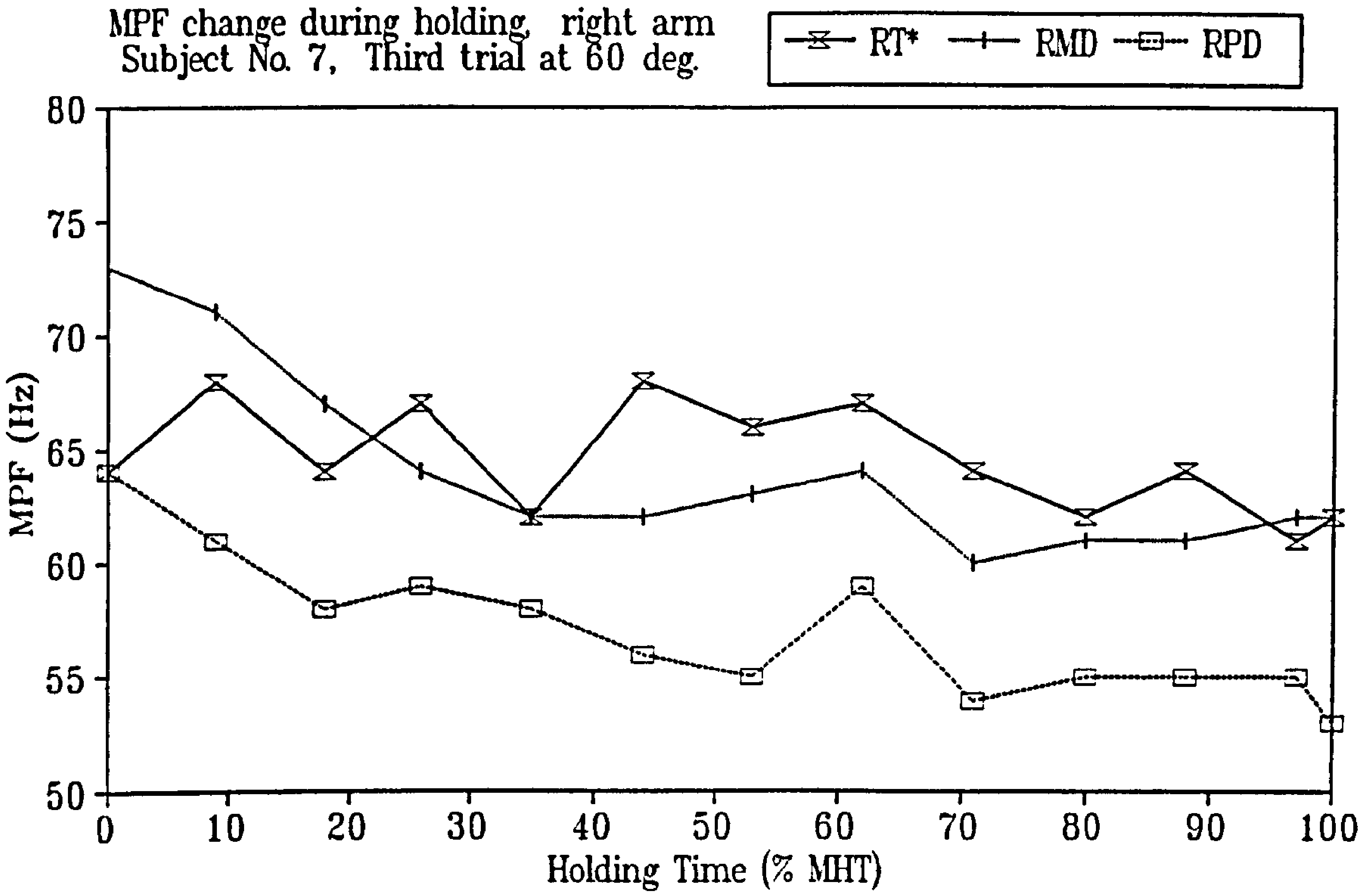


Figure D.46 Changes in MPF for subject No. 6 at 60 deg.



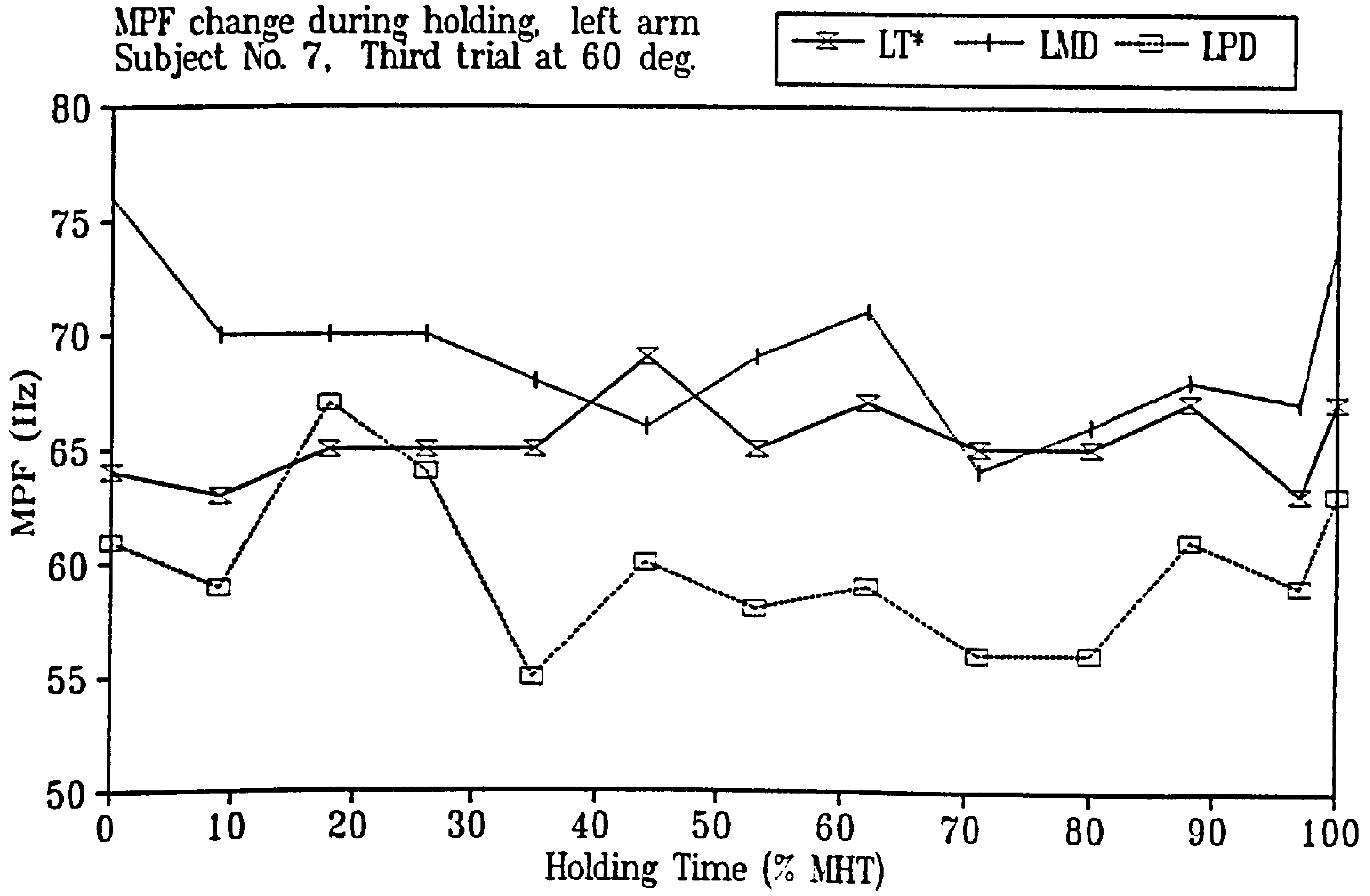
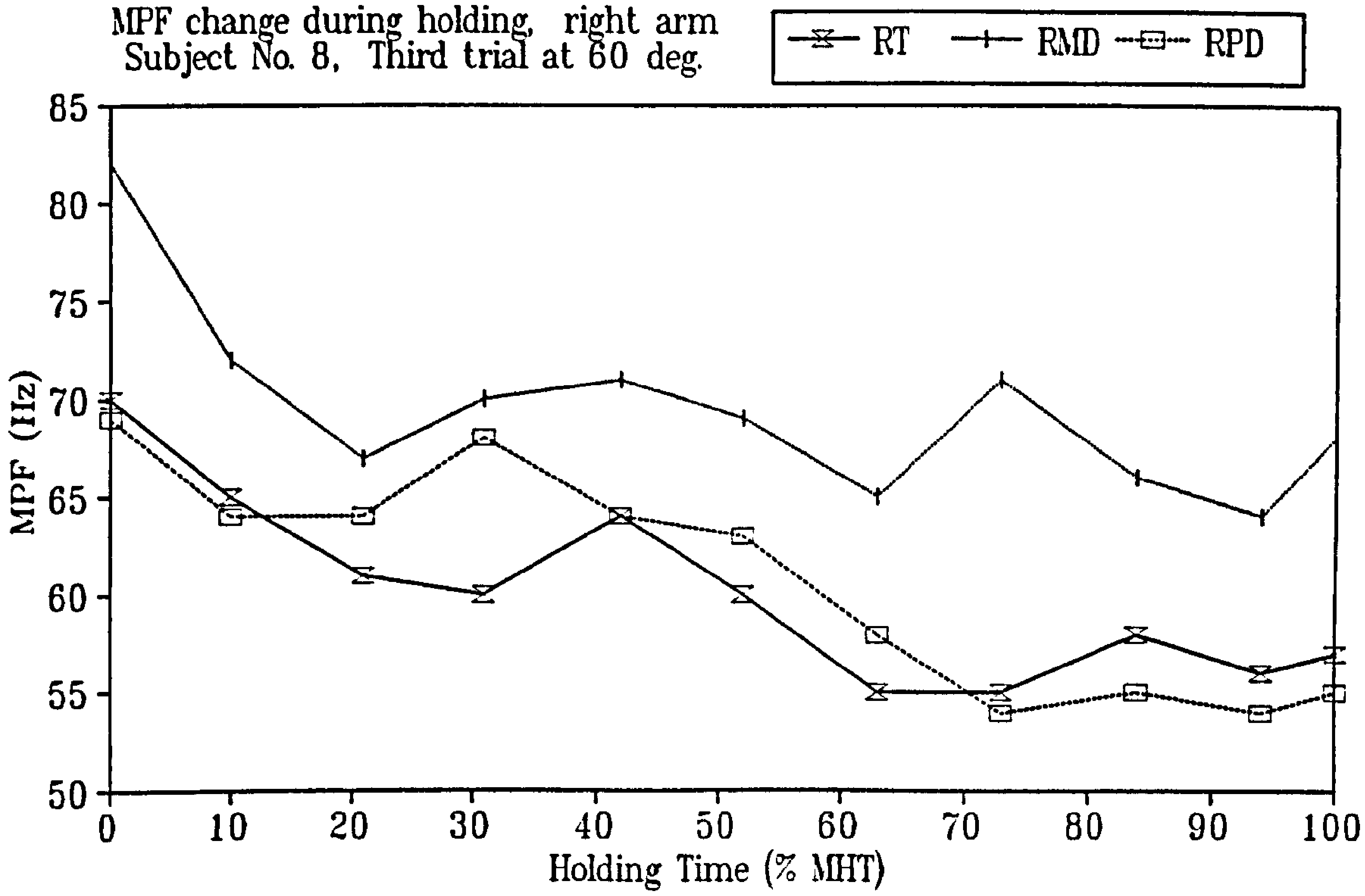


Figure D.47 Changes in MPF for subject No. 7 at 60 deg.



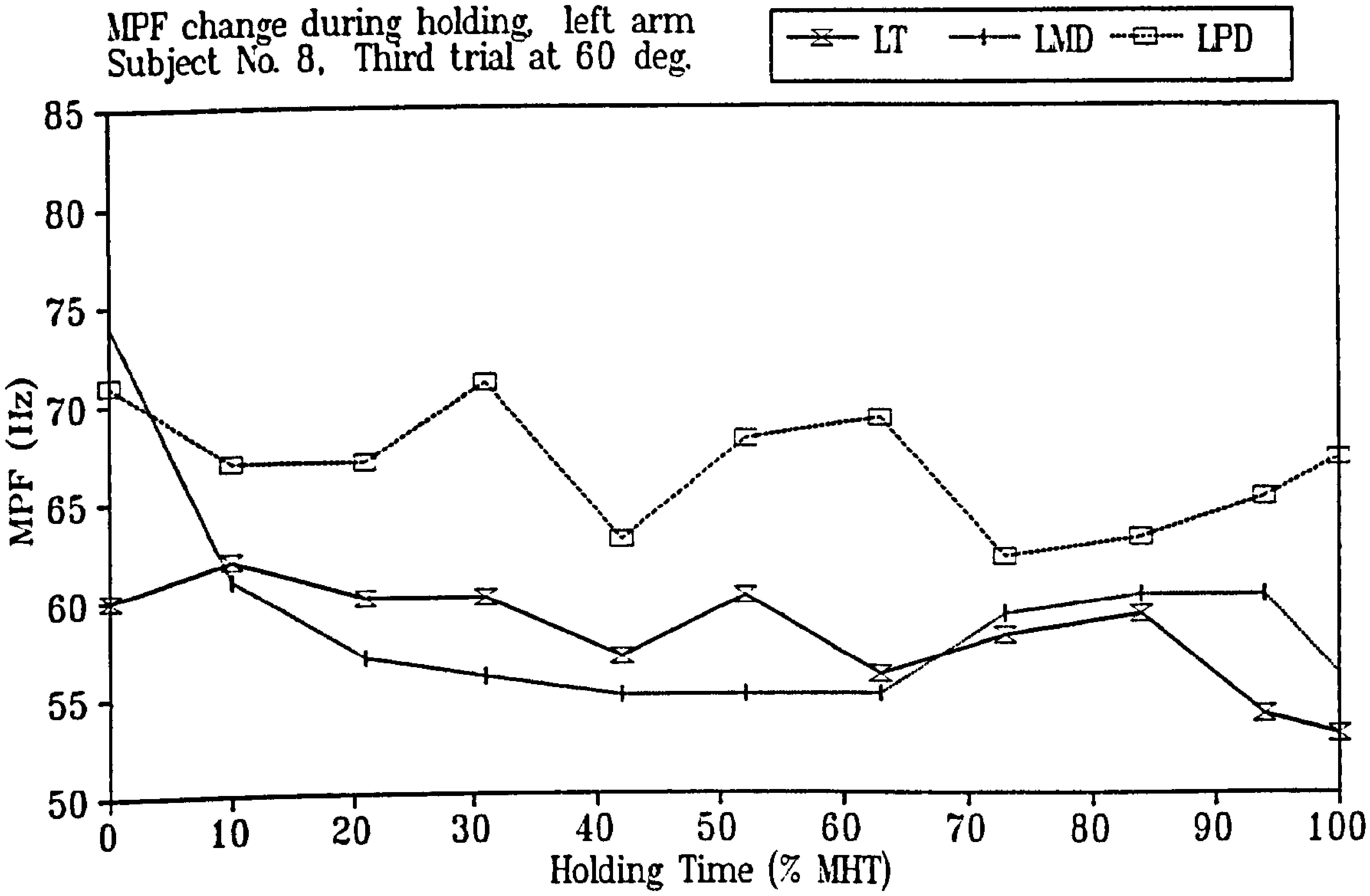
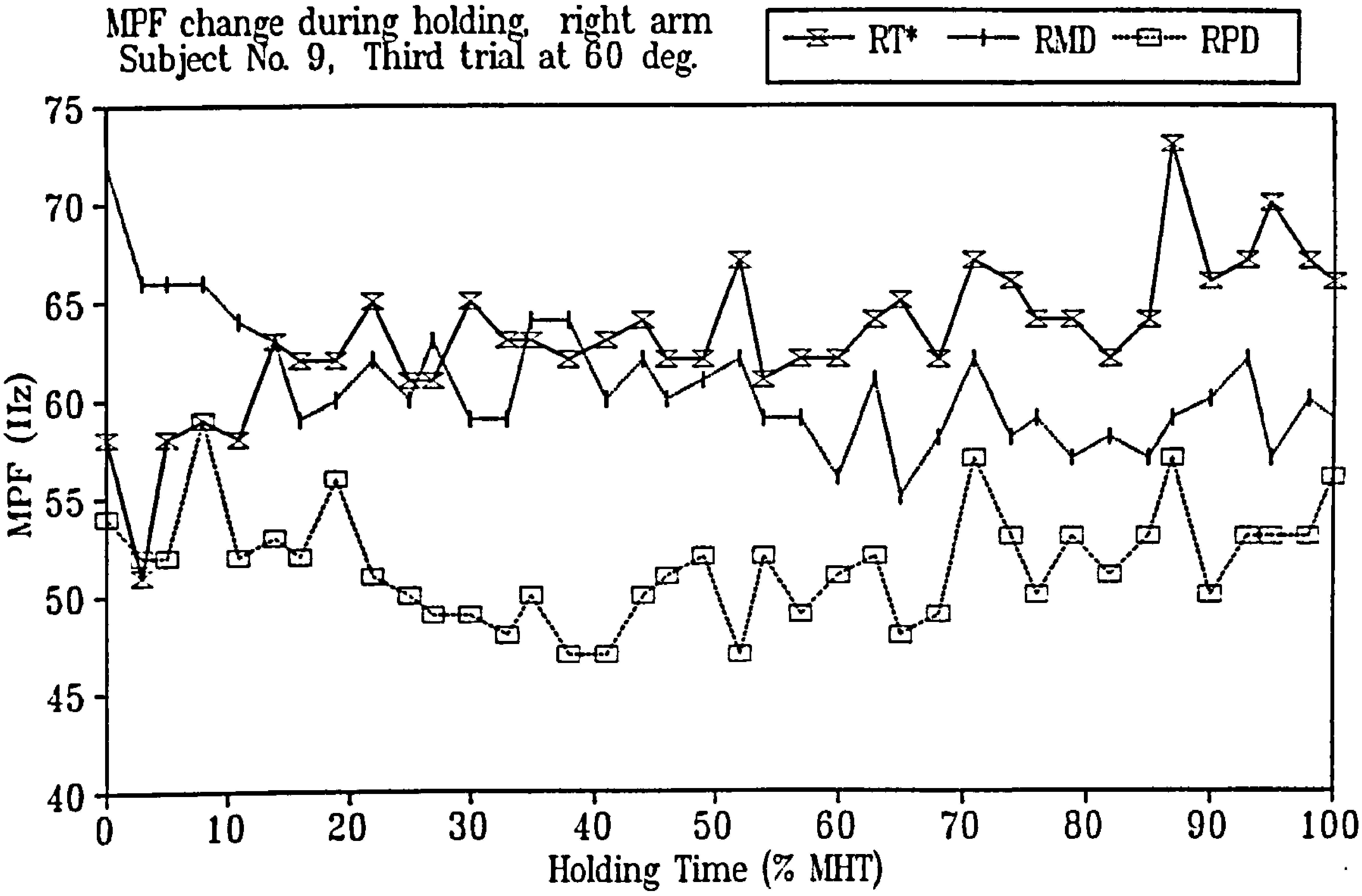


Figure D.48 Changes in MPF for subject No. 8 at 60 deg.



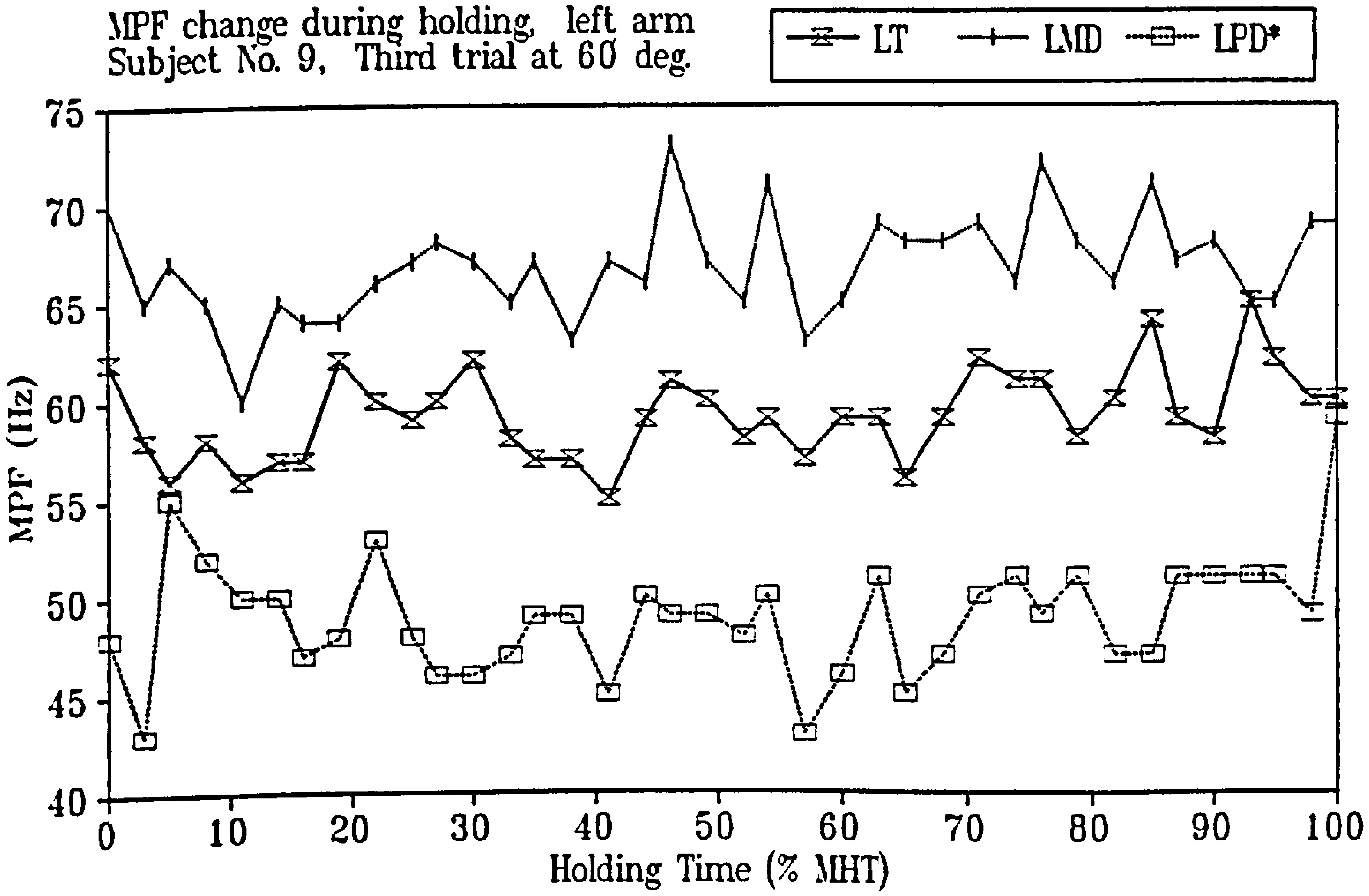
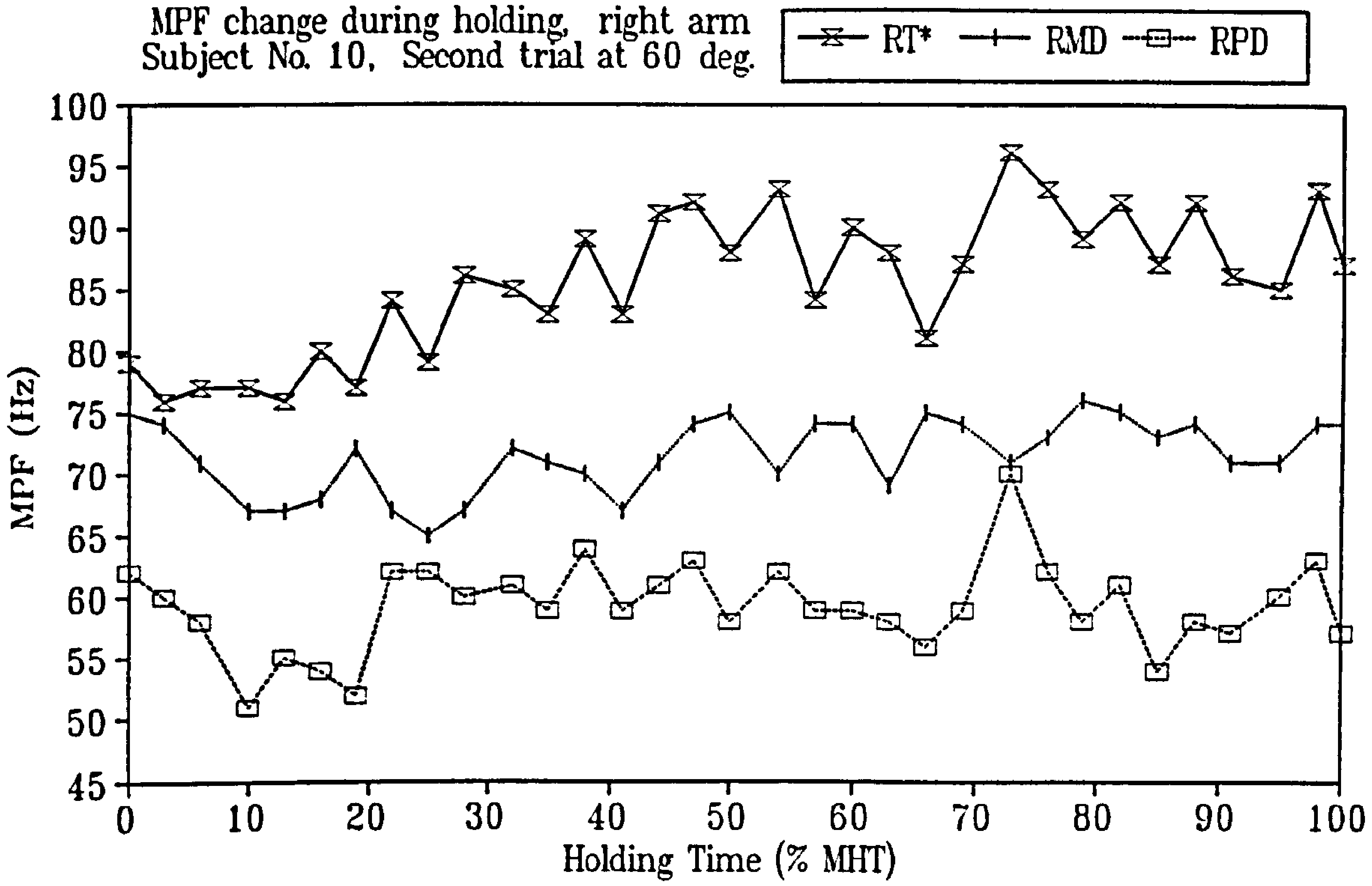


Fig D.49 Changes in MPF for subject No. 9 at 60 deg.



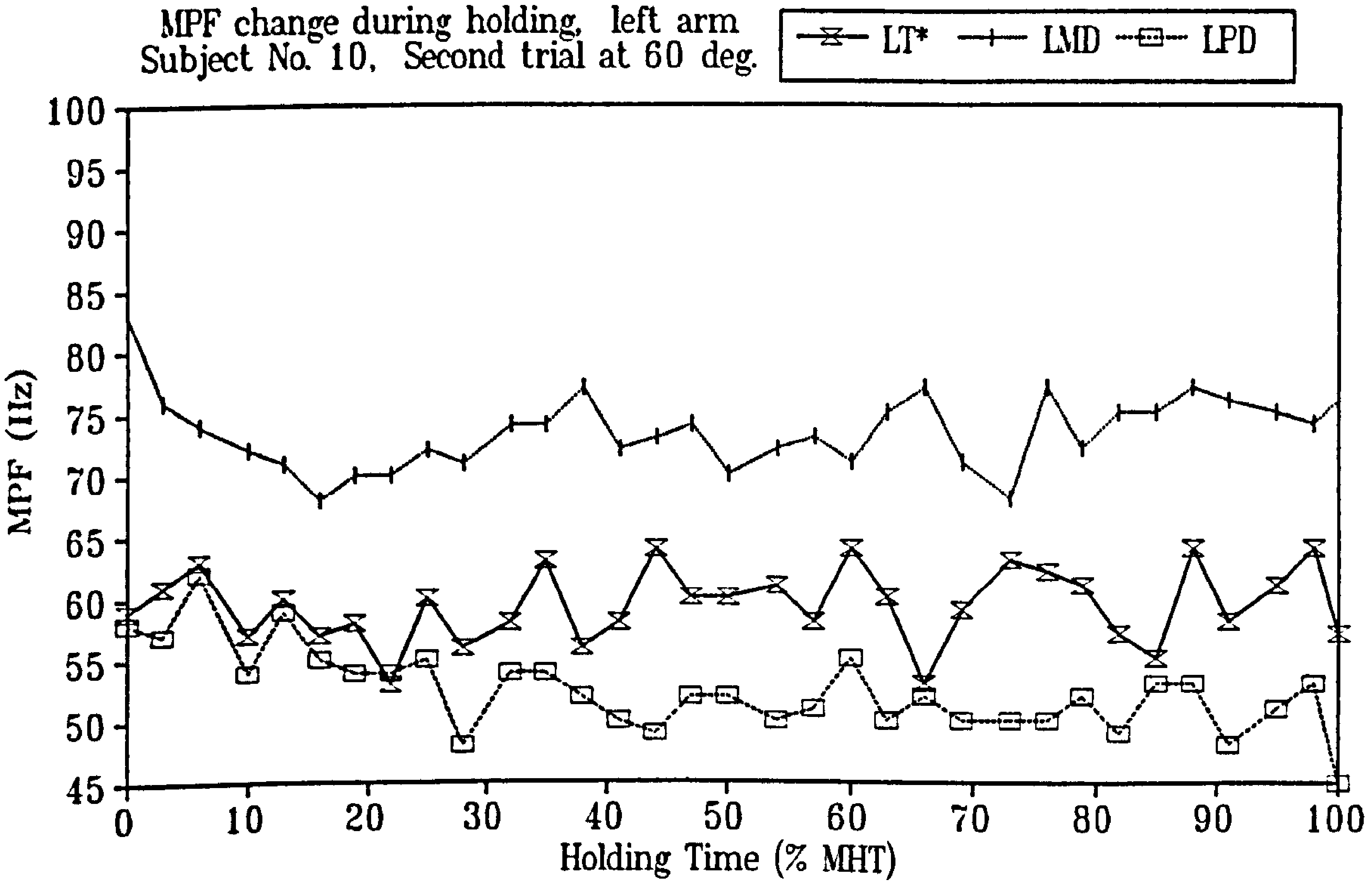
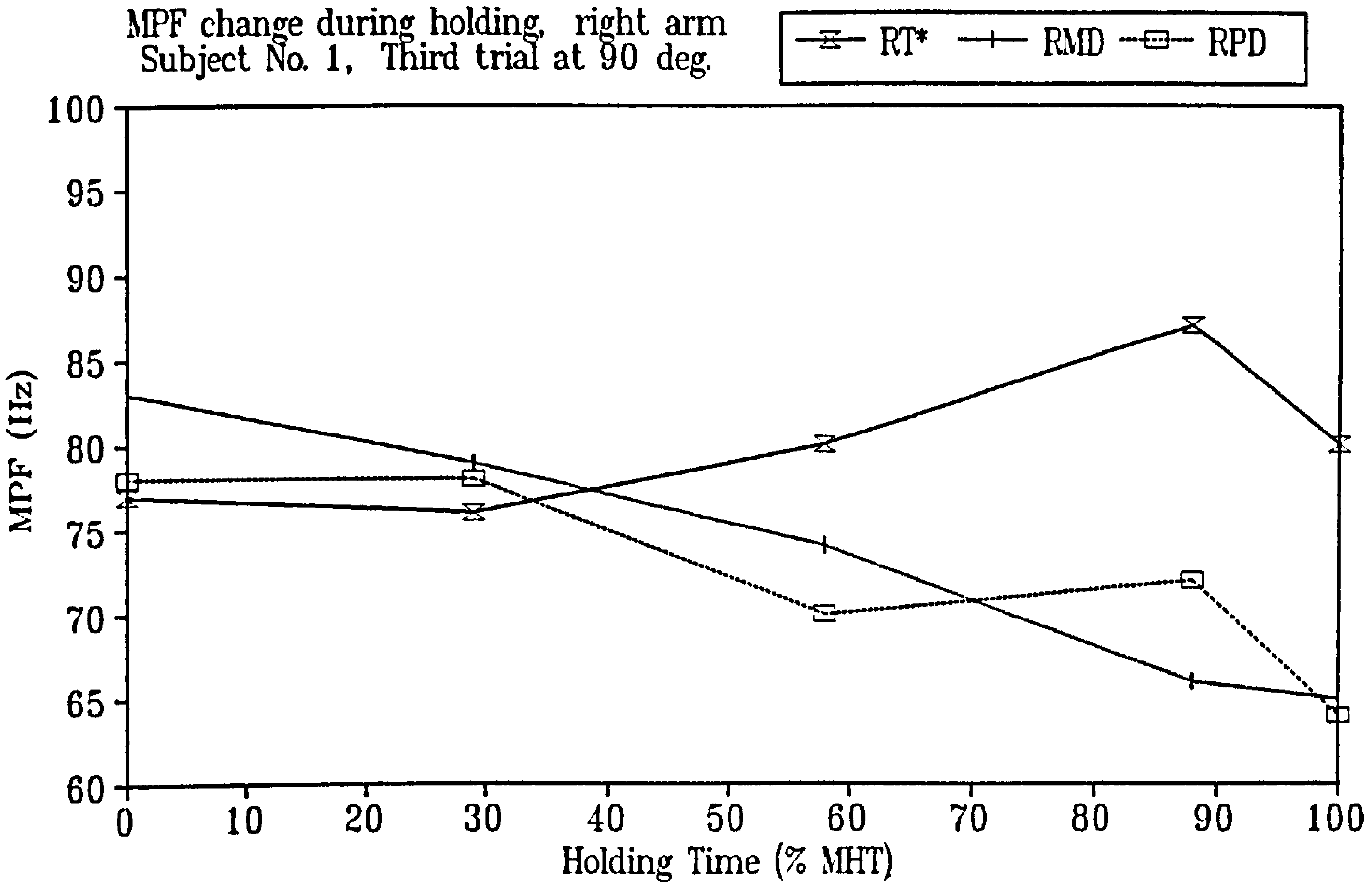


Figure D.50 Changes in MPF for subject No. 10 at 60 deg.



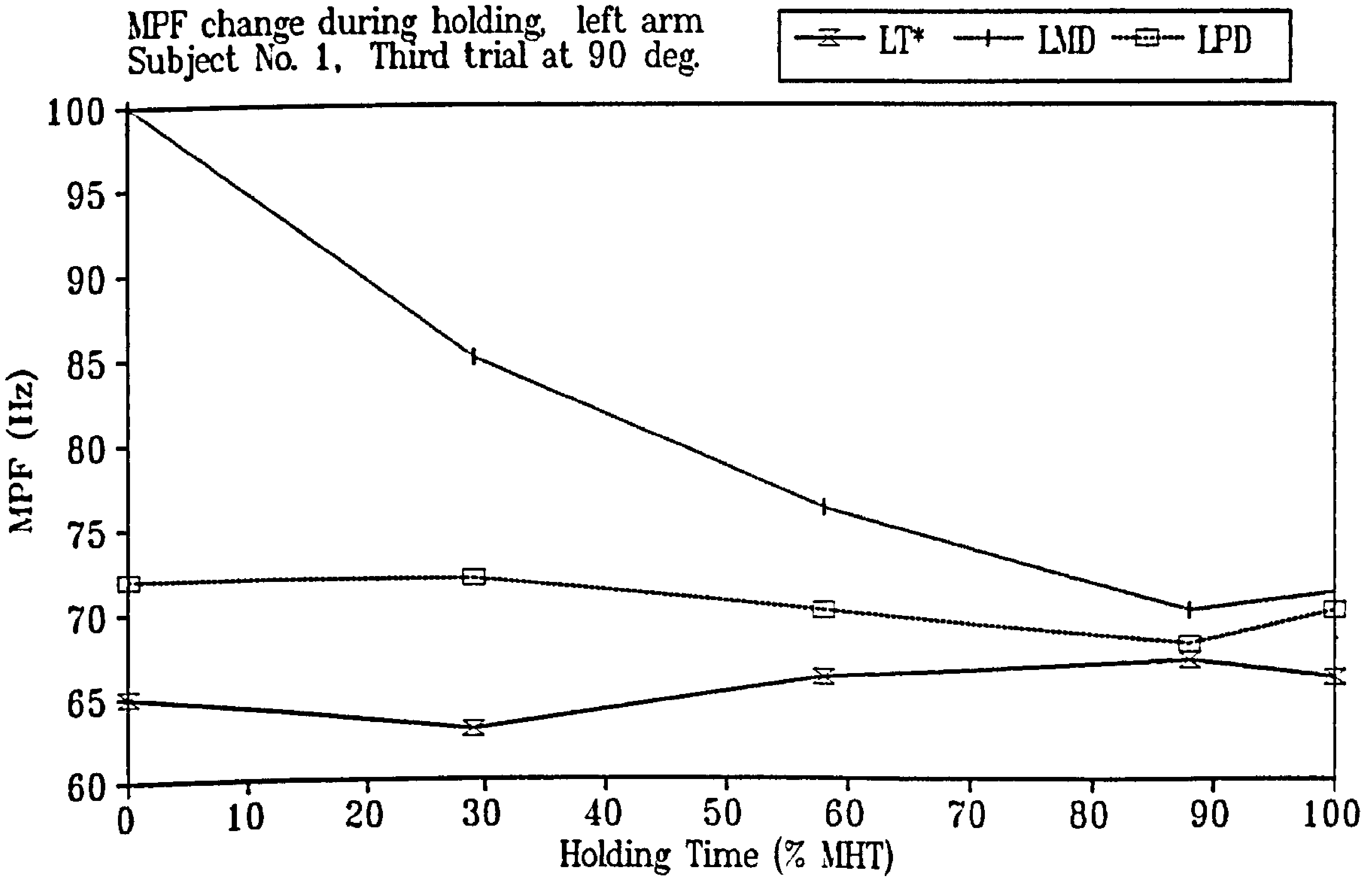


Figure D.51 Changes in MPF for subject No. 1 at 90 deg.

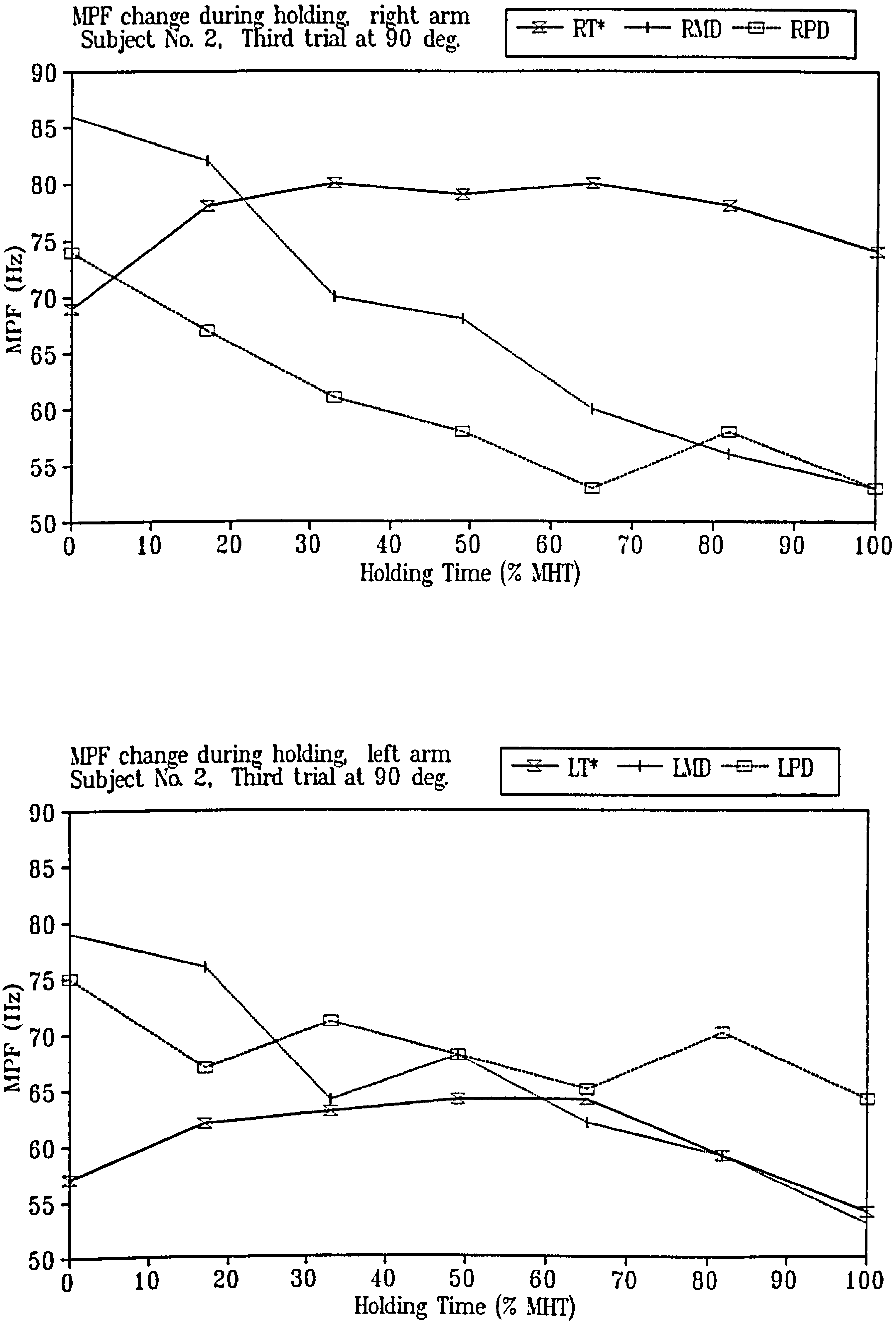
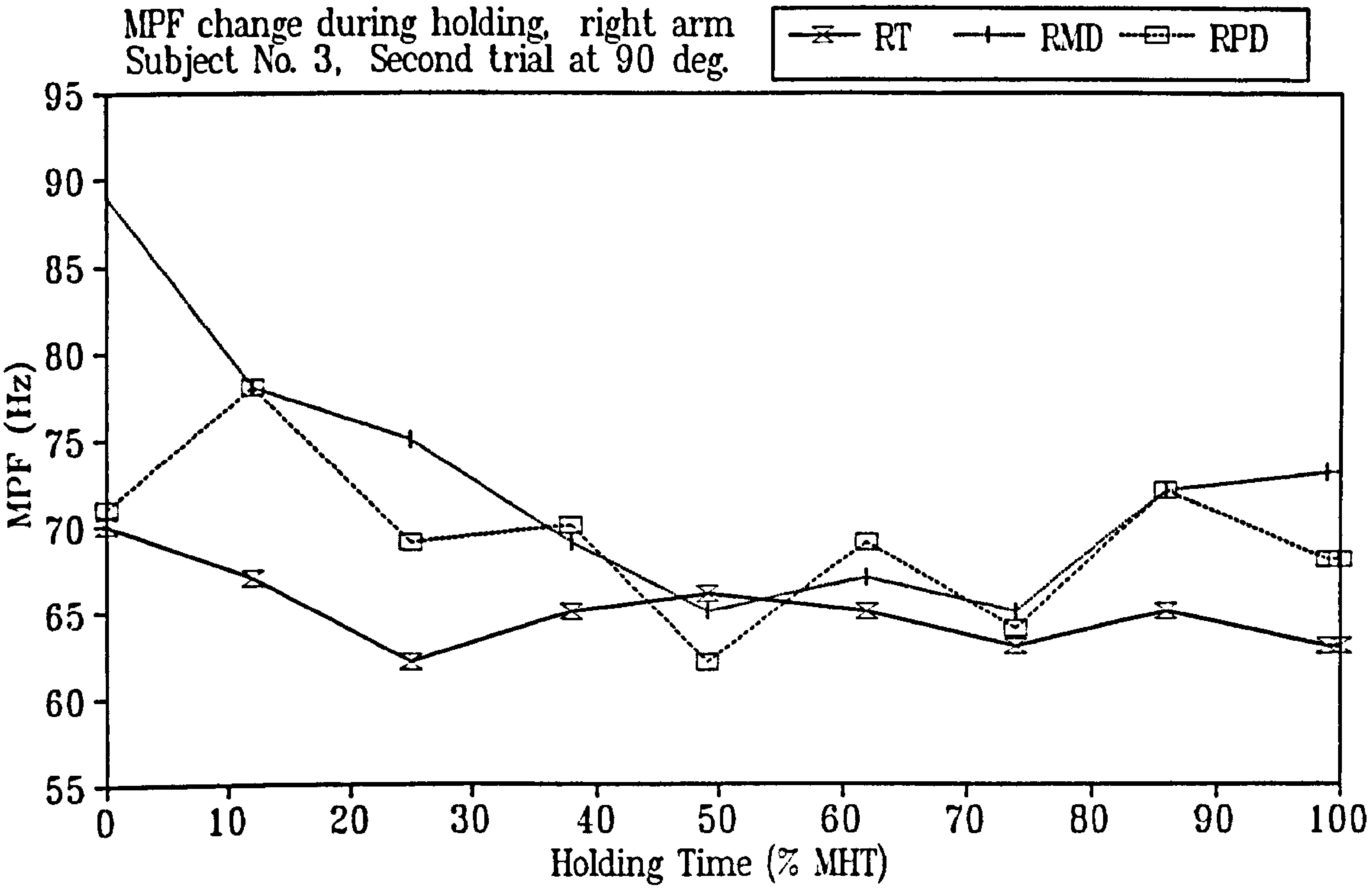


Fig D.52 Changes in MPF for subject No. 2 at 90 deg.



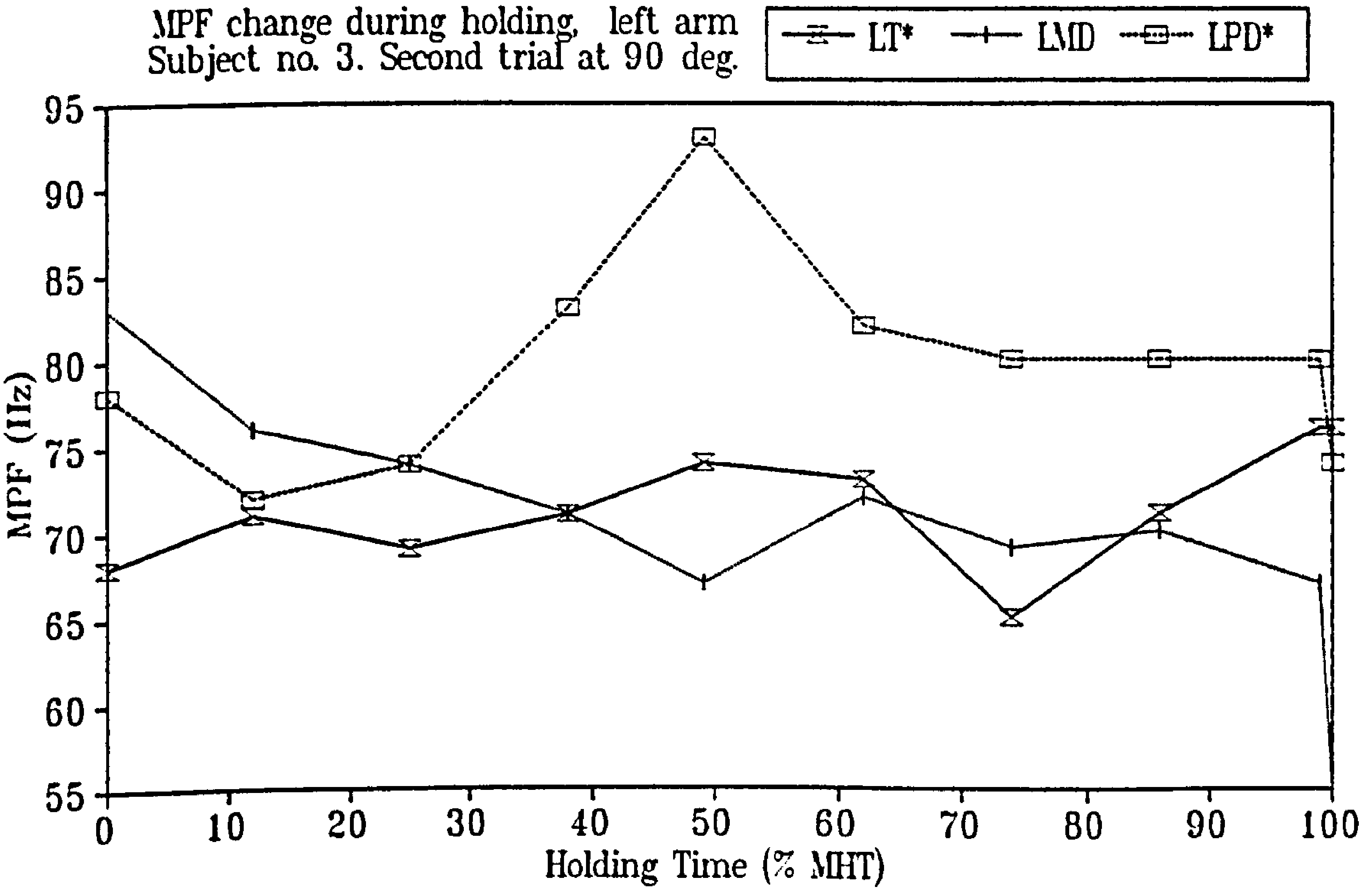
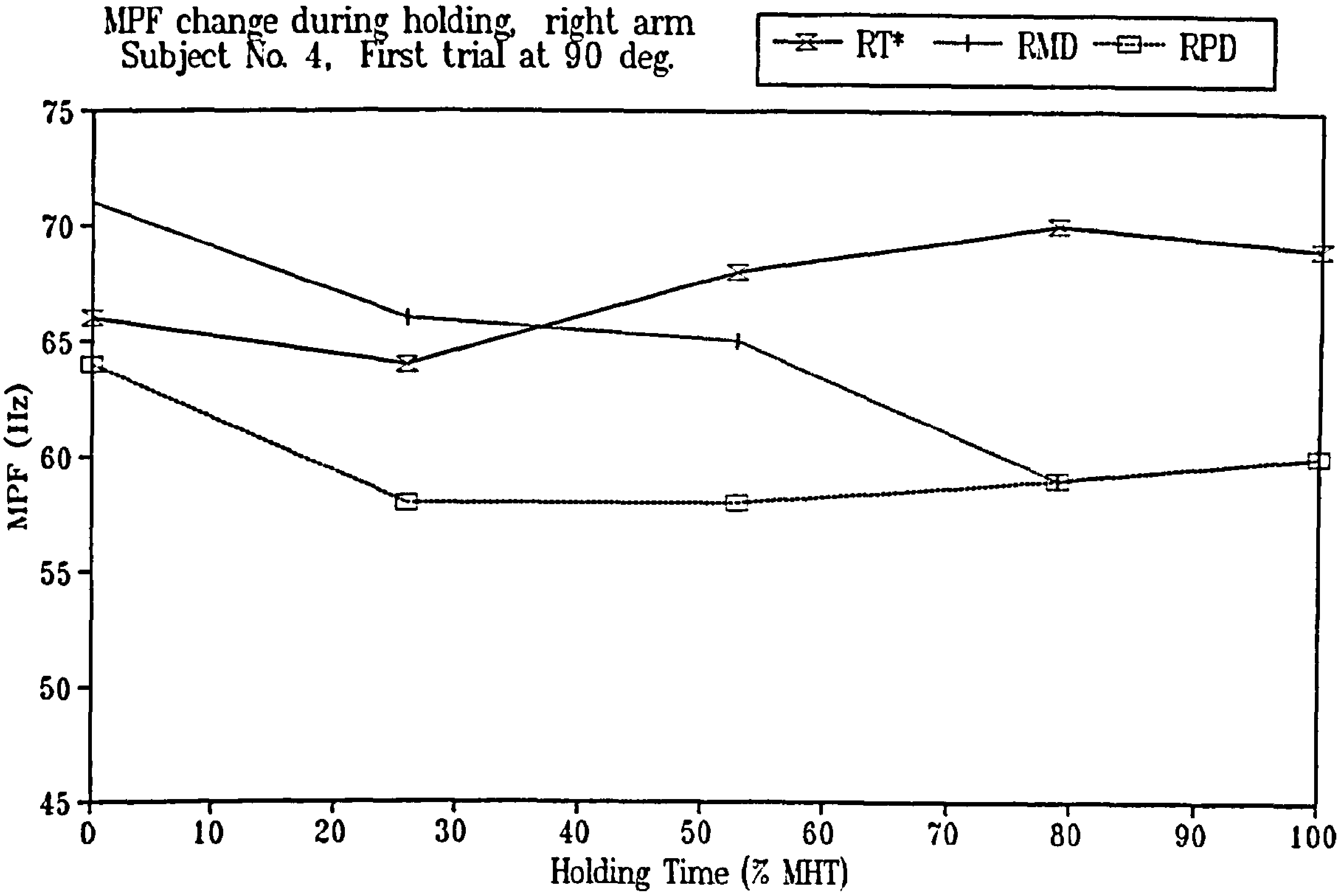


Figure D.53 Changes in MPF for subject No. 3 at 90 deg.



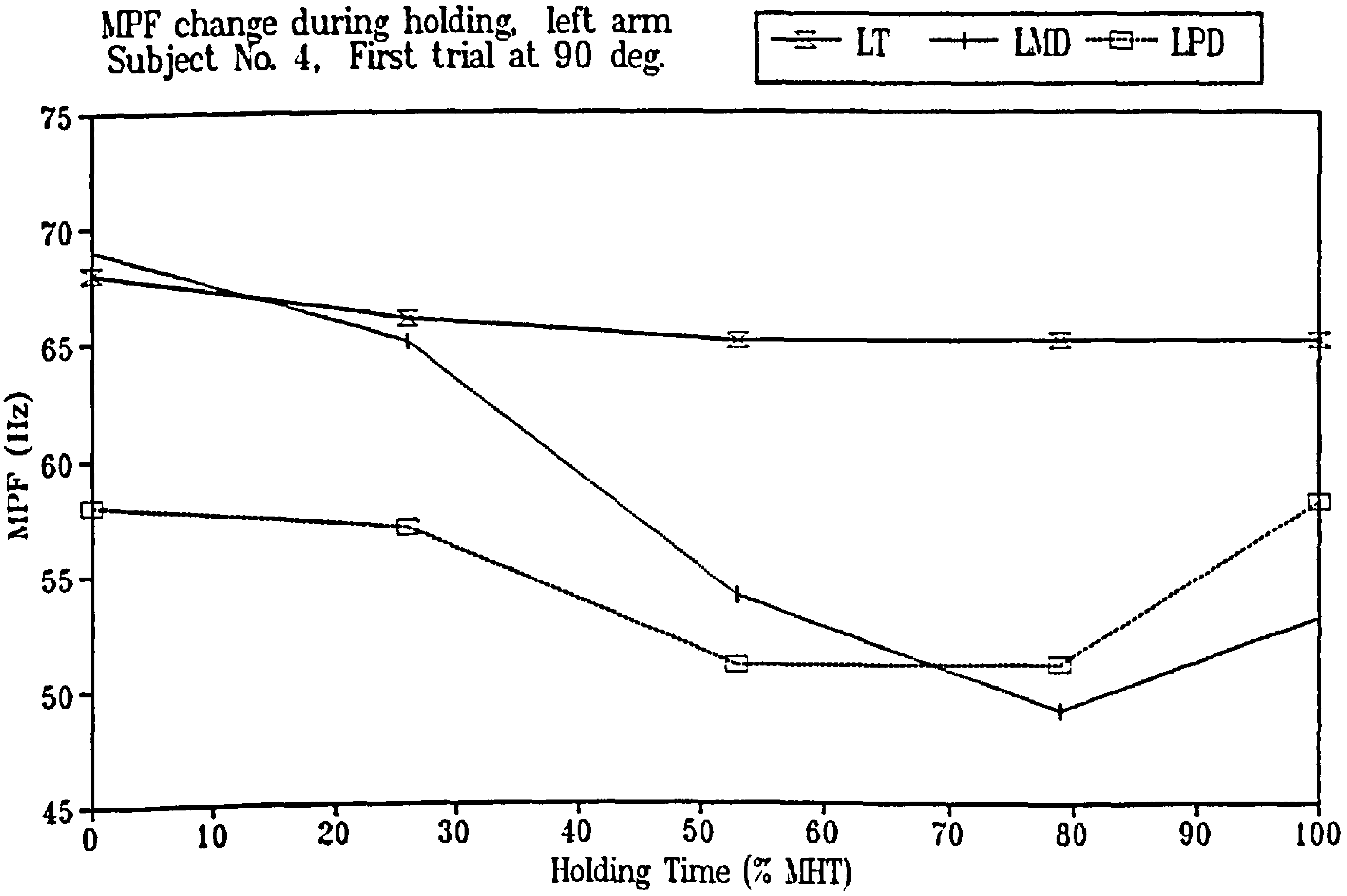
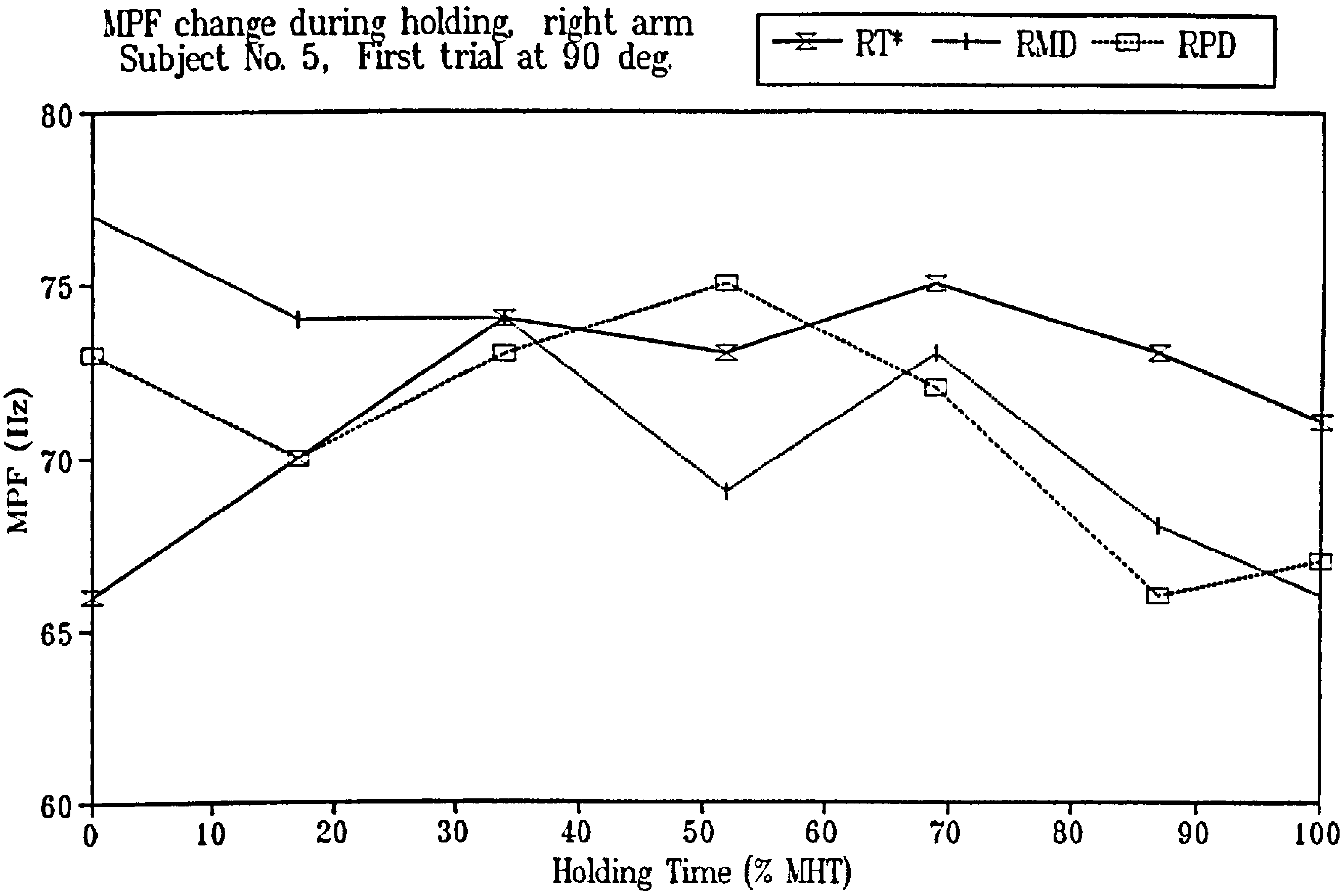


Figure D.54 Changes in MPF for subject No. 4 at 90 deg.



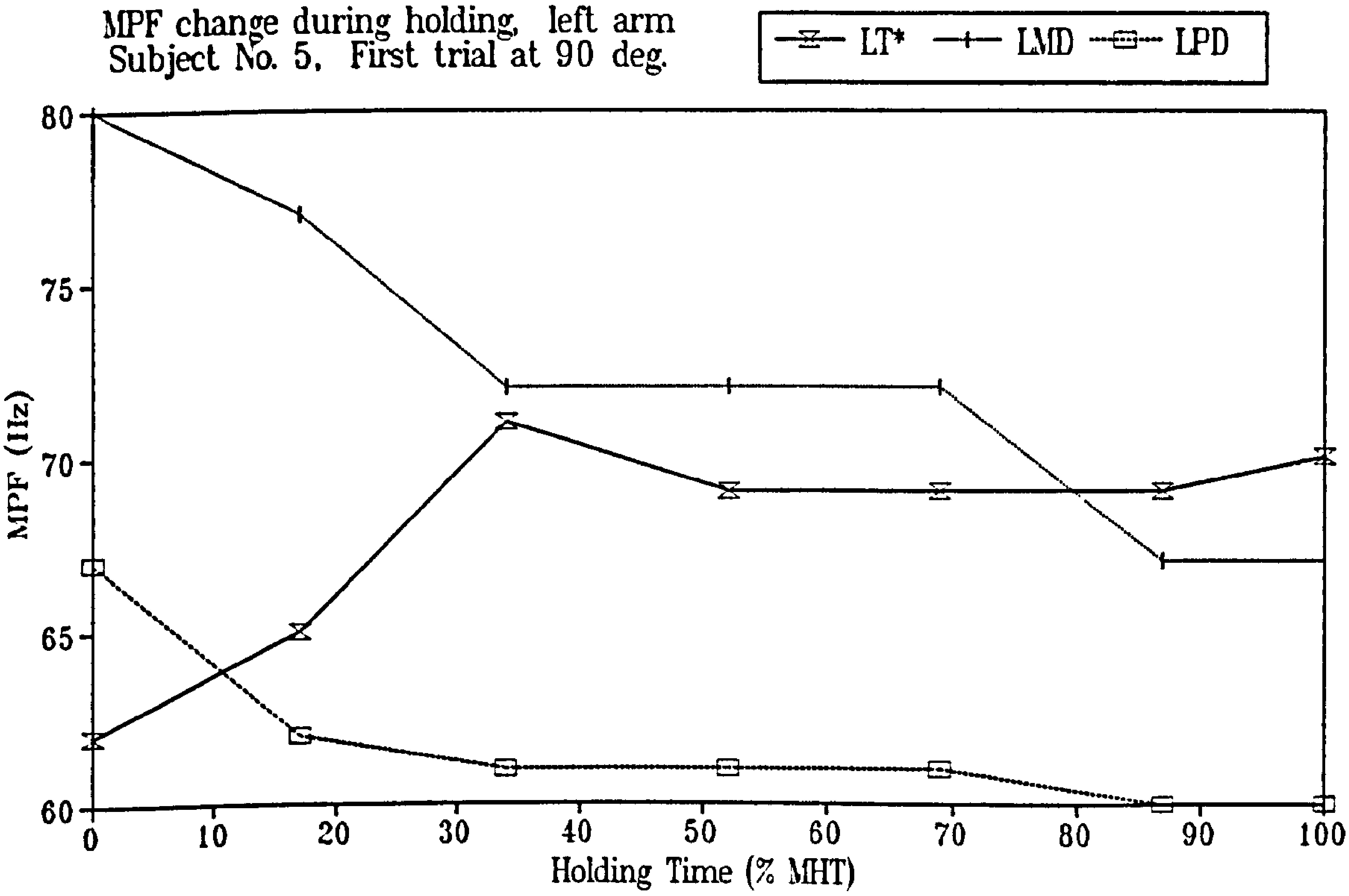
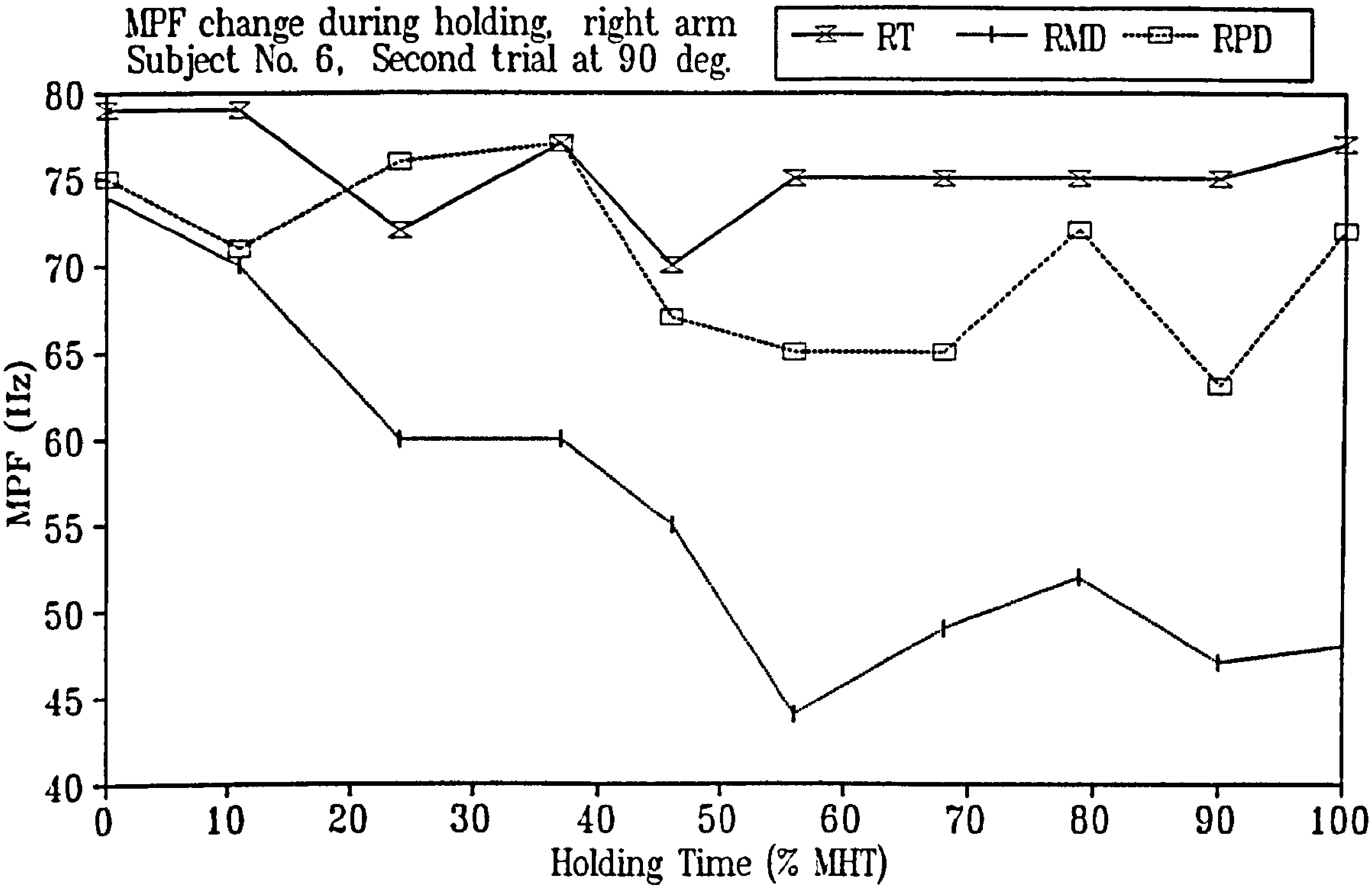


Figure D.55 Changes in MPF for subject No. 5 at 90 deg.



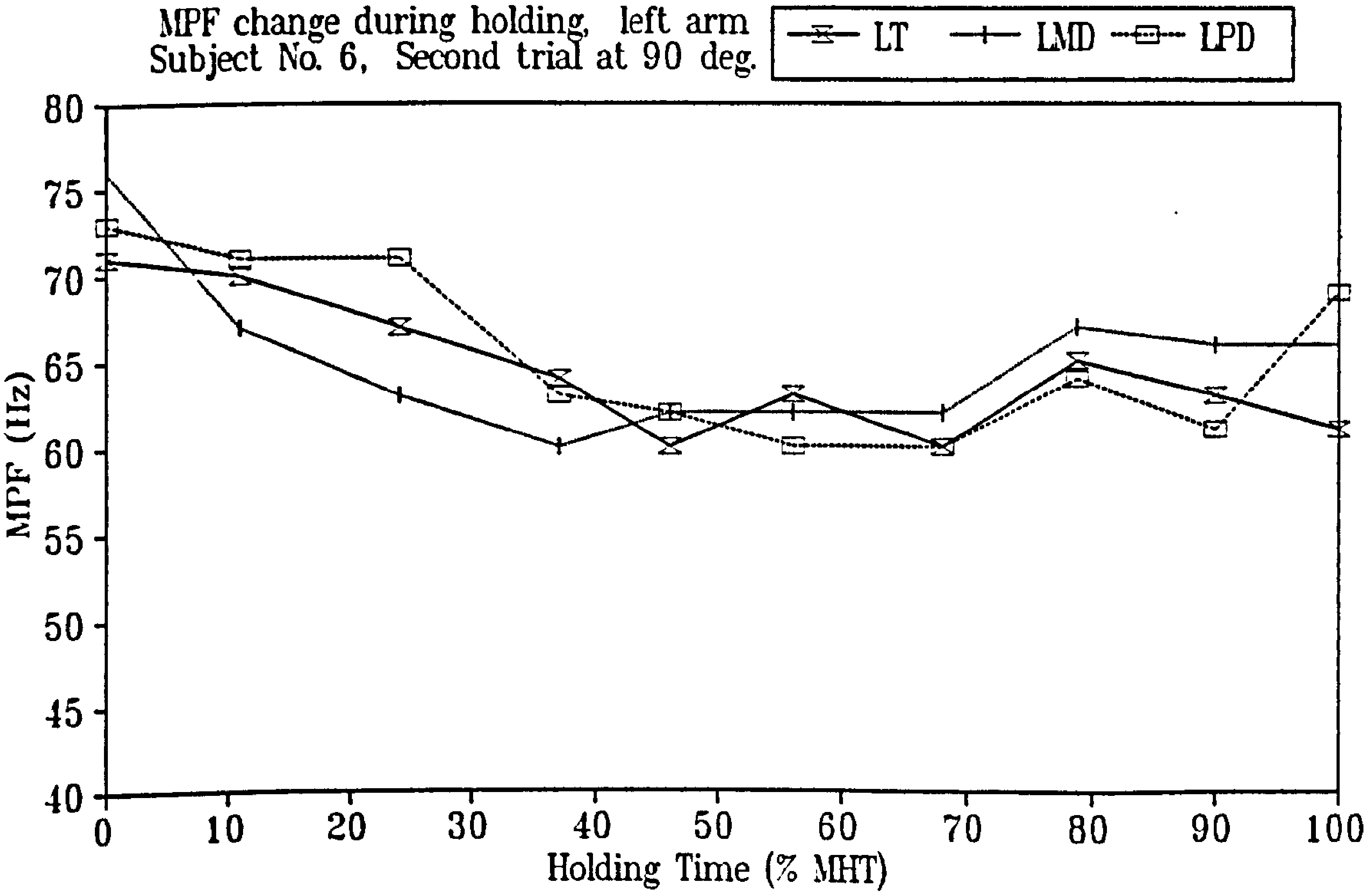
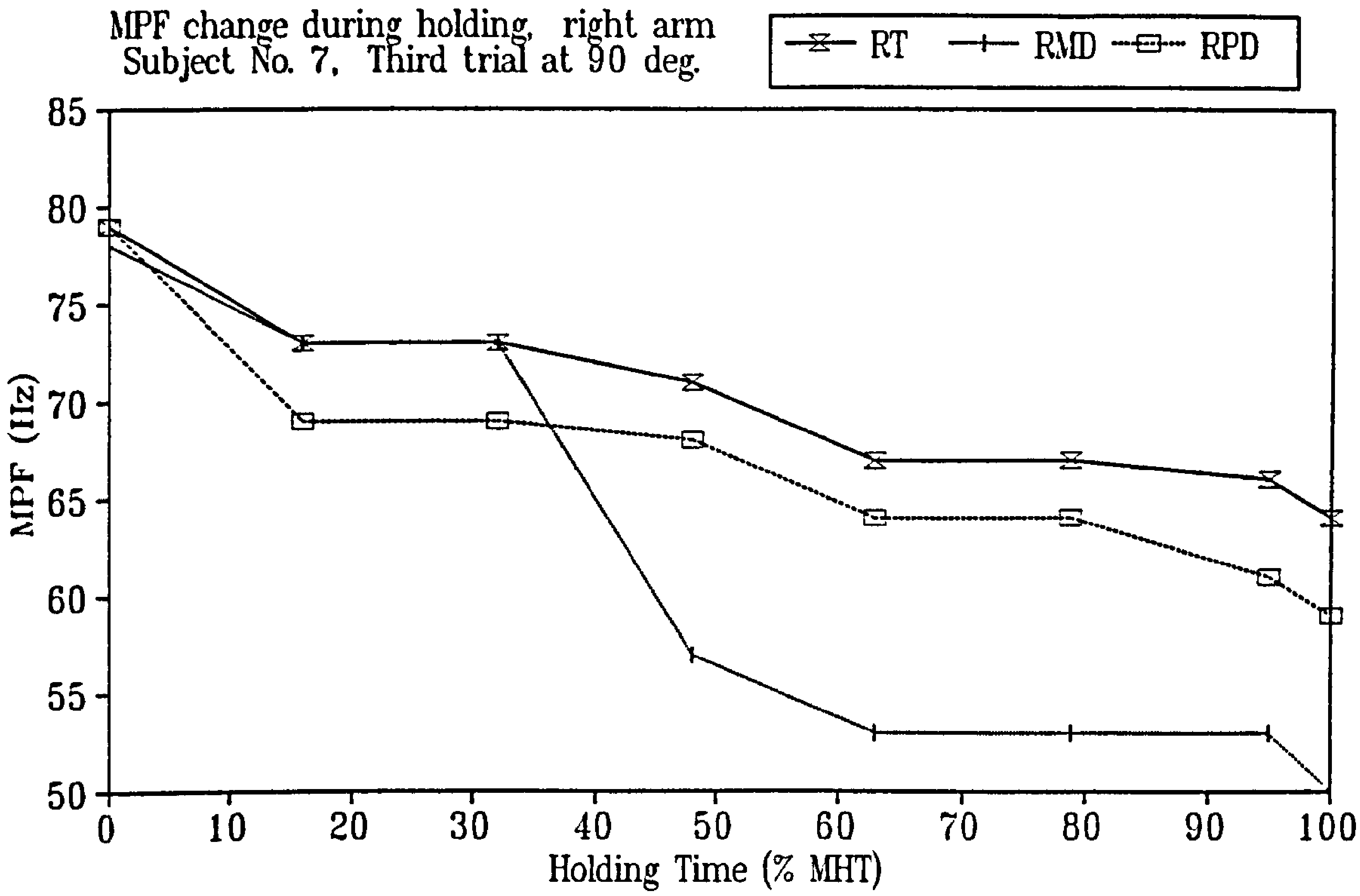


Figure D.56 Changes in MPF for subject No. 6 at 90 deg.



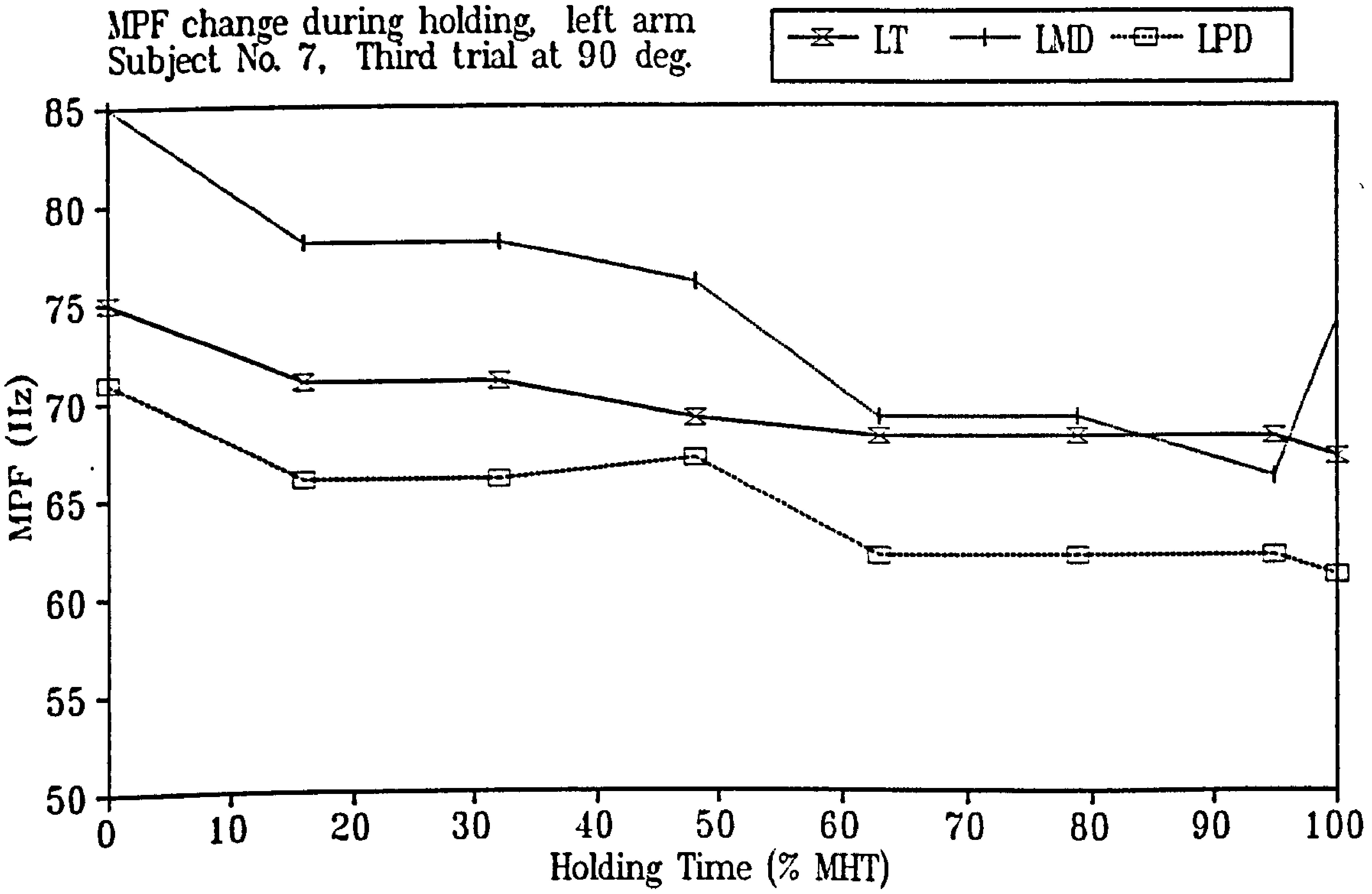


Figure y.57 Changes in MPF for subject No. 7 at 90 deg.

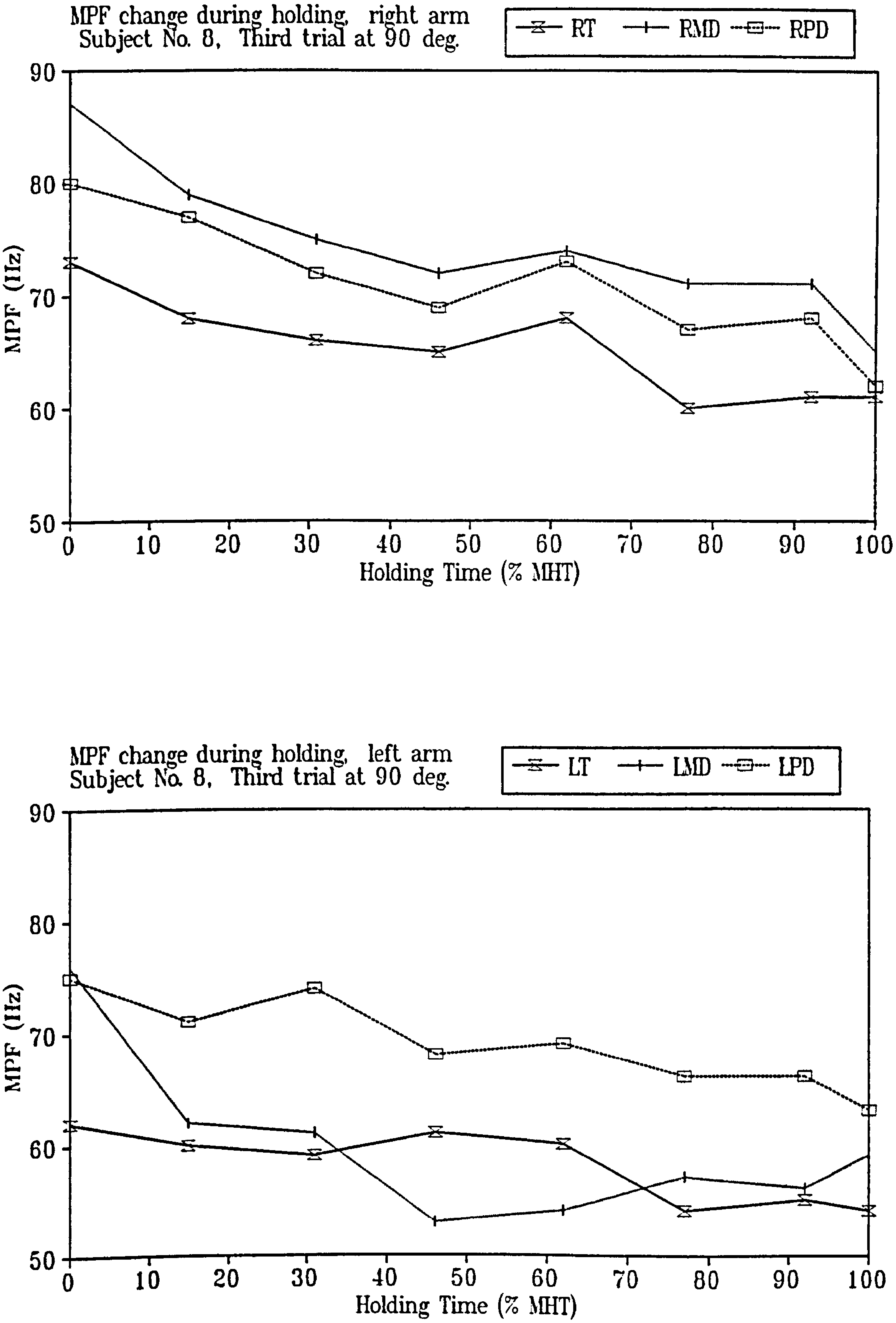
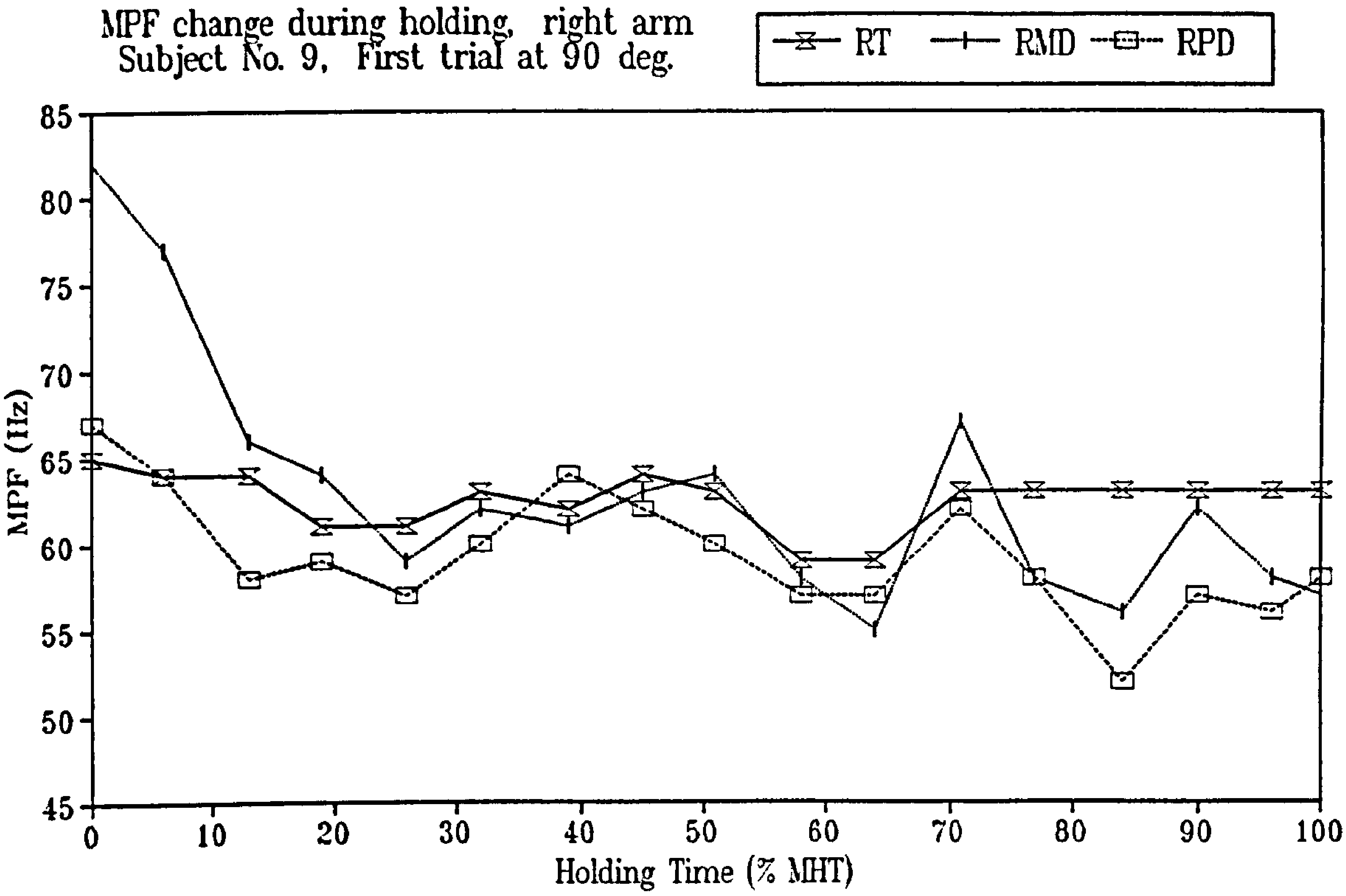


Figure D.58 Changes in MPF for subject No. 8 at 90 deg.


MPF change during holding, right arm
Subject No. 9, First trial at 90 deg.

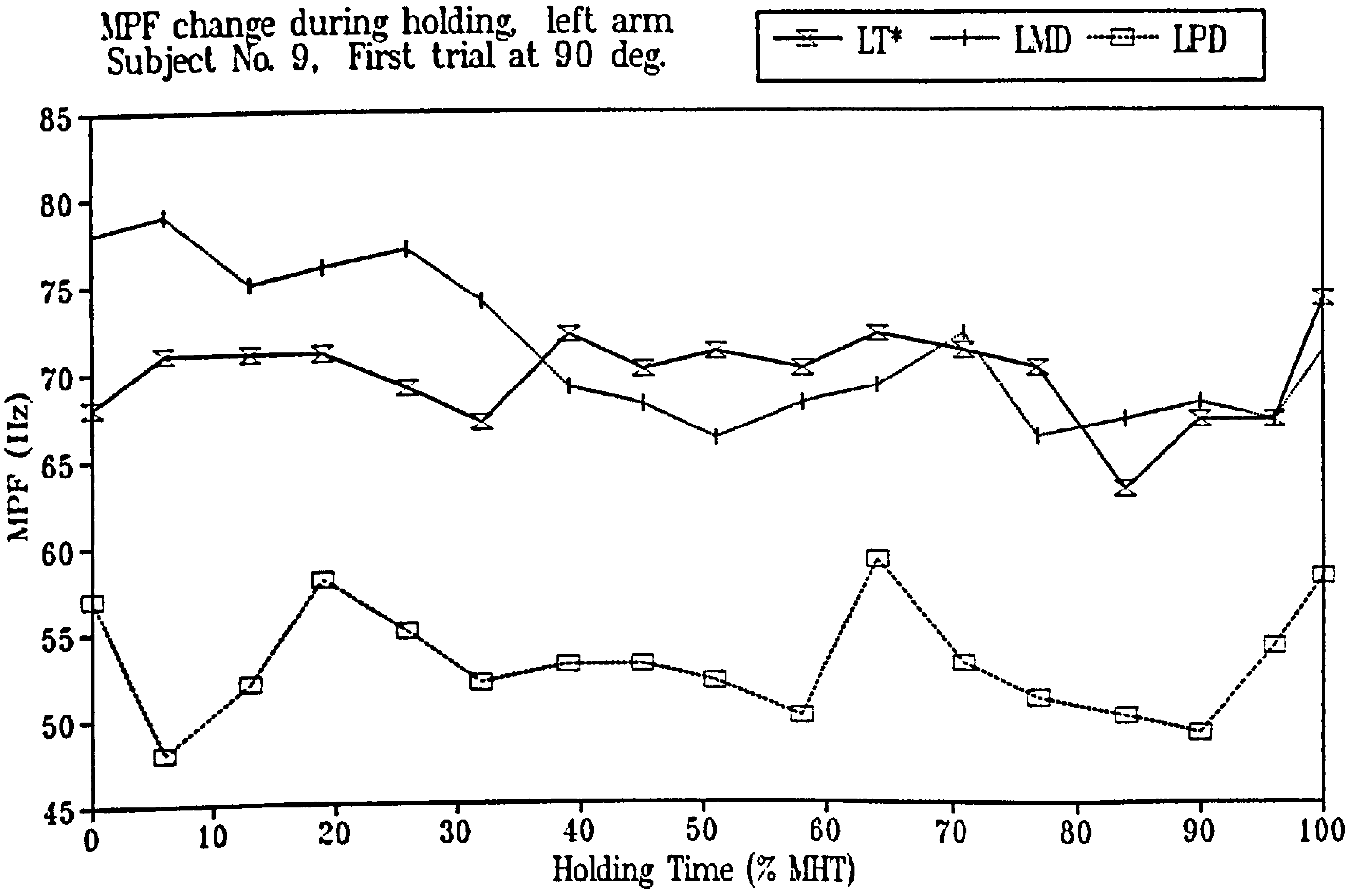
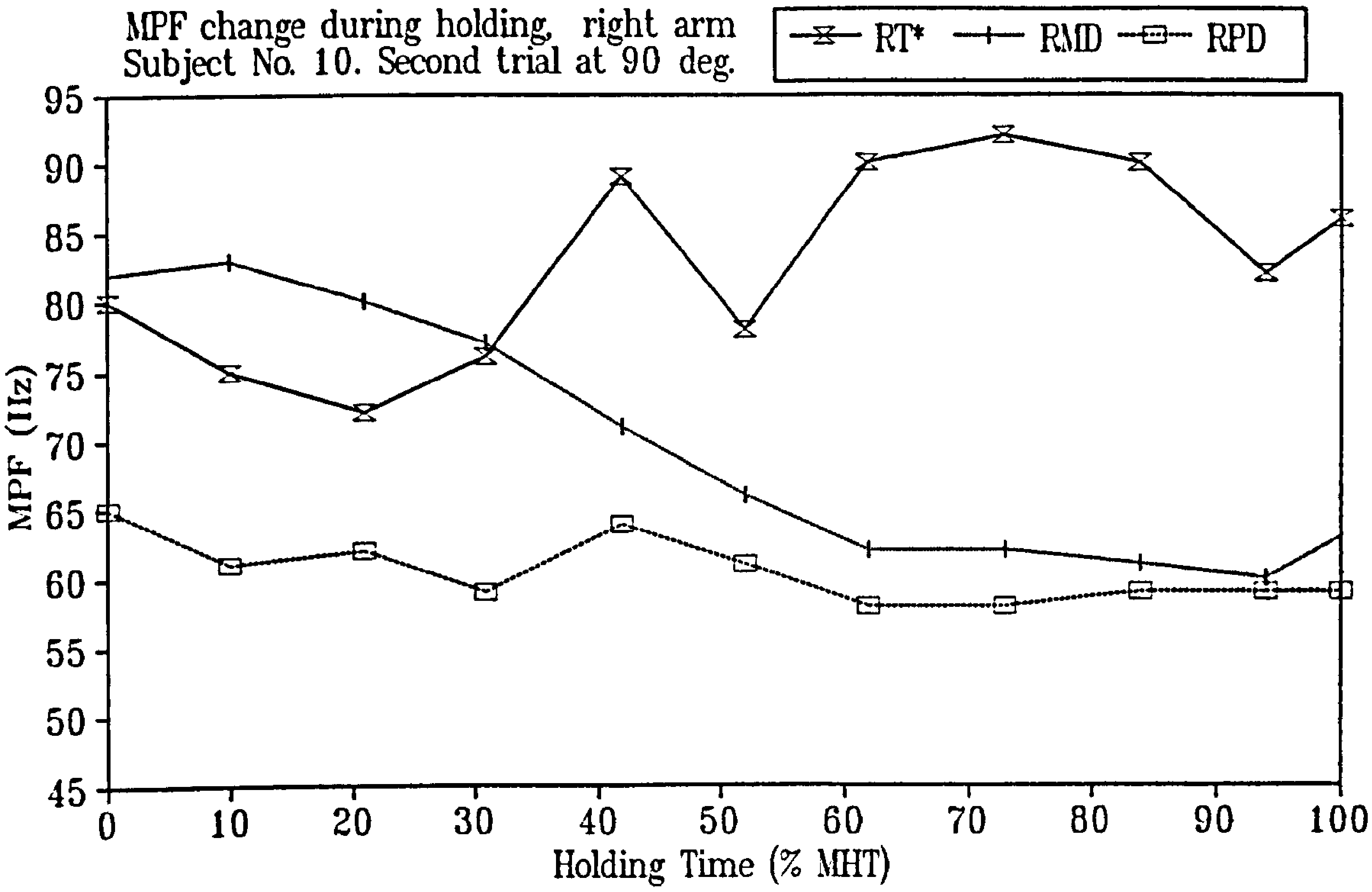

MPF change during holding, left arm
Subject No. 9, First trial at 90 deg.

Figure D.59 Changes in MPF for subject No. 9 at 90 deg.



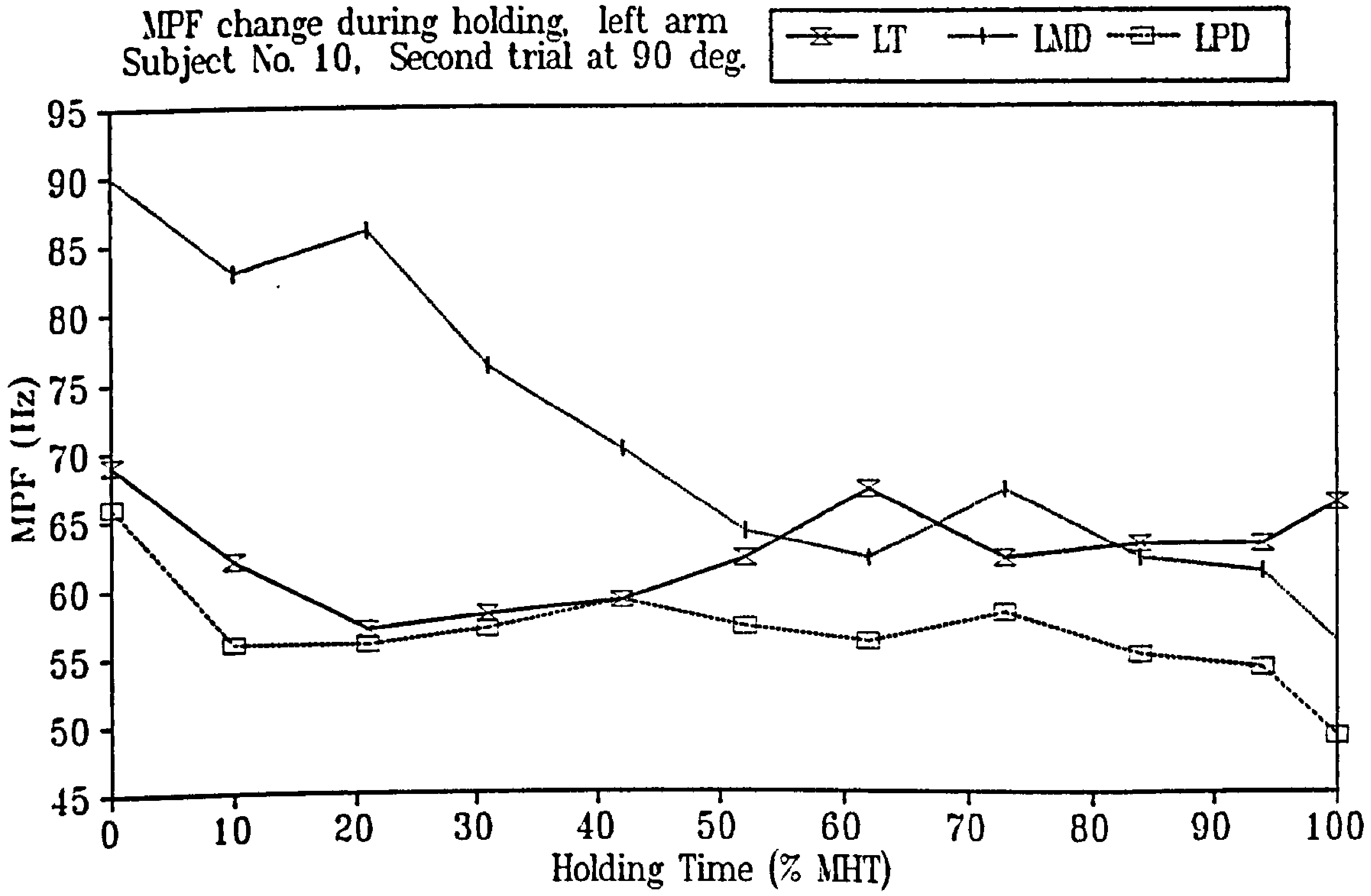


Figure D.60 Changes in MPF for subject No. 10 at 90 deg.

Appendix E

Publications generated in the course of the study

Serratos-Pérez J.N., Haslegrave C.M. (1992)
Modelling fatigue and recovery in working postures
Contemporary Ergonomics 1992, pp 66-71.

Serratos-Pérez J.N., Haslegrave C.M. (1993)
Monitoring fatigue development from static
postures taxing the shoulder region.
The Ergonomics of Manual Work, pp 261-264.

Serratos-Pérez J.N., Haslegrave C.M. (1994)
Relationship between subjective perception and EMG signs
of muscular fatigue in shoulder-loading postures.
Accepted for presentation at the 12th IEA Congress

PAGE/PAGES
EXCLUDED
UNDER
INSTRUCTION
FROM
UNIVERSITY