School of Physiotherapy

The Assessment of Hip Abductor Activity Using Ultrasound Imaging

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Statement of Originality

To the best of my knowledge and belief this thesis contains no material previously published by any other person except where due acknowledgement has been made. This thesis contains no material which has been accepted for the award of any other degree or diploma in any university.

Angela Dieterich, January 2013

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Abstract

Hip pain prevalence is between 12-19% in older Caucasian populations. Hip joint impairments are often explained by aberrations in joint shape that cause detrimental joint mechanics. Little attention has been paid to the active control of joint mechanics. Altered motor patterns may increase local joint loading and shear, both mechanisms which contribute to joint pain and structural damage. The hypothesis of this thesis states that non-traumatic, anterior hip pain is associated with altered muscular hip joint control.

Approximately 90% of human muscles, including most hip muscles, are deep lying and assessable only with invasive electromyography. The application of fine-wire electromyography is ethically and clinically limited. Alternatively, ultrasound imaging has been applied increasingly to assess deep muscles activity, but to date not to hip muscles. This research was devoted to the exploration of ultrasound imaging for assessing activity level and onset of the gluteus medius and minimus muscles, the main hip abductors and stabilizers in the frontal plane.

A series of six studies was conducted to compare measurements of muscle thickness and onset of muscle motion by B-mode, M-mode, Pulsed-Wave Doppler ultrasound, surface and fine-wire electromyography; to develop a computed measurement protocol; and to assess gluteus medius and minimus activity patterns in subjects with hip pain compared to controls.

Study One indicated good intra-tester reliability of B-mode and M-mode ultrasound measurements of muscle thickness, $ICC_{3,k} > 0.90$ within-day and >0.80 between days. The study indicated a unique advantage of M-mode ultrasound for the visual control of deep muscle relaxation.

Study Two identified the longitudinal scanning plane as being valuable for ultrasound measurements on the hip abductors because the changing muscle shape during contraction affected the transducer stability less than in transverse scanning.

Study Three demonstrated that while gluteus medius thickened by 21% during high-level isometric activity, gluteus minimus thickness was constant, not allowing for estimations of the activity level by ultrasound measurements of muscle thickness change.

Study Four indicated the superiority of M-mode ultrasound compared to Pulsed-Wave Doppler for comparing activity-related muscle motion in adjacent muscles. In healthy subjects, a sequential onset of gluteus minimus activating 36 ms before gluteus medius was found. In spite of excellent correlation of electromyography and ultrasound measured activity onsets, the precision of estimating excitation onset from muscle motion was unexpectedly low, mean difference >125 ms, limits of agreement >220 ms. In an additional experiment, technical limitations in ultrasound data synchronization could be identified.

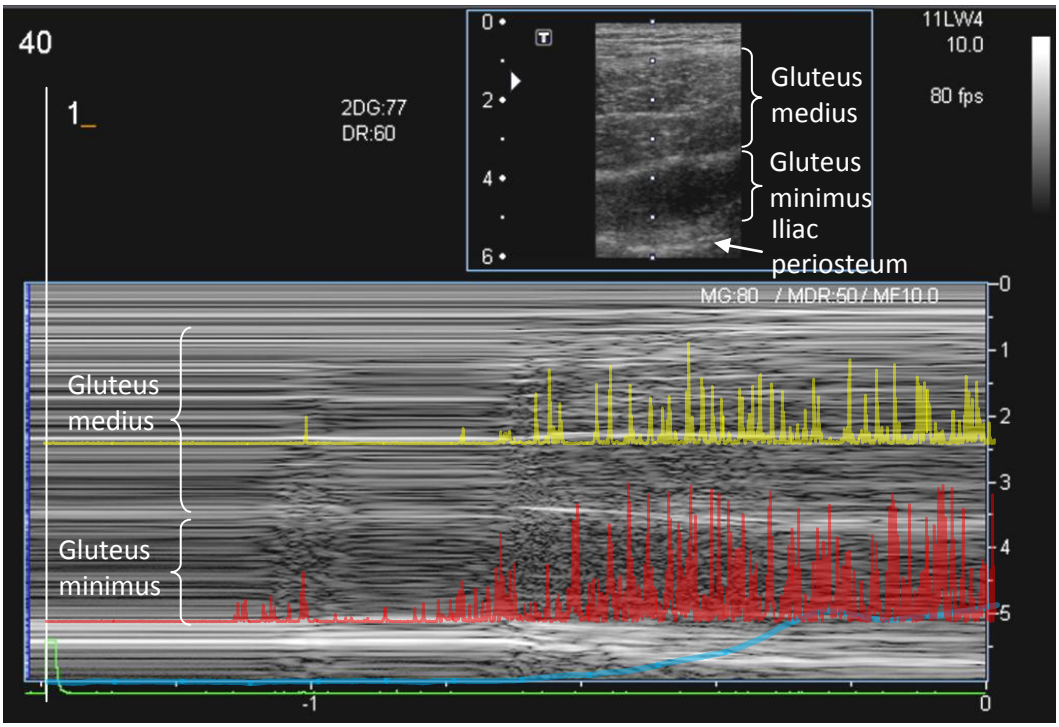
Study Five applied newly developed computed pattern detection, based on the Teager-Kaiser Energy Operator, in the M-mode ultrasound traces of activity-related muscle motion. The study demonstrated that the high-frequency components of M-mode traces of muscle activity are most closely related to muscle excitation, mean difference 21 ms, limits of agreement 90 ms. These results were limited to control subjects. Measurements on four subjects with hip pain indicated that the relationship between excitation and muscle motion may be affected by pain.

Study Six was a clinical study using M-mode ultrasound in which the activity patterns of the hip abductor muscles of 35 participants with non-traumatic anterior hip pain were compared to those of 35 matched controls. During step-down onto the painful limb, subjects with pain demonstrated a significantly (p 0.001) earlier onset, by 103 ms, of high-frequency greyscale patterns in the gluteus minimus muscle. The result supports the hypothesis of the thesis.

This doctoral research explored the application of M-mode ultrasound imaging for rehabilitative purposes. A new measurement of activity-related muscle motion based on high-frequency greyscale patterns was developed and the clinical relevance of this measurement was demonstrated. M-mode ultrasound enabled non-invasive measurements of mechanical aspects of muscle activity which indicated altered motor patterns in subjects with hip pathology. The observation and measurement of activity-related muscle motion by M-mode ultrasound may enhance the understanding of the coordination of superficial and deep muscles and provide new insights into muscular contributions to pathology and their therapy.

One picture is worth a thousand words. Fred R. Barnard (Advertising manager, Printers' Ink, 10 March 1927)

Photography, as a powerful medium of expression and communication, offers an infinite variety of perception, interpretation and execution. Ansel Adams (Photographer, 1902 – 1984)



M-mode ultrasound trace of muscle motion of the gluteus medius and minimus muscles during isometric hip abduction, including initial hesitation; overlying, the concurrently recorded electromyography signals of the deep gluteus medius (yellow) and gluteus minimus (red). The blue trace is the resultant force measured by dynamometry.

Conference Presentations and a Grant to Continue this Research

Physiokongress 18. – 20.06.2009, platform presentation; Fellbach / Stuttgart, Germany.
Title: 'Werden der tiefe und der oberflächliche Gluteus medius gleichzeitig aktiviert?' (Do the deep and the superficial gluteus medius activate synchronously?) A. Dieterich, Dr. C. Pickard, G. Strauss, Dr. G.T. Allison, L. Deshon

HAWK Hochschule für Angewandte Wissenschaft und Kunst, 8.7.2009, presentation and talk; Hildesheim, Germany.
Title: 'Werden der tiefe und der oberflächliche Gluteus medius gleichzeitig aktiviert?' (Do the deep and the superficial gluteus medius activate synchronously?) A. Dieterich, Dr. C. Pickard, G. Strauss, Dr. G.T. Allison, L. Deshon

WCPT Congress 2011 (World Confederation for Physical Therapy), 20. - 23.06.2011, platform presentation; Amsterdam, Netherlands.
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Myograd, International Research Training Group for Myology, Berlin/Paris, Summer School 13. – 17.06.2011, poster presentation; Berlin, Germany.
Title: 'M-mode ultrasound of hip abductor activation' Dieterich A, Pickard C, Deshon L, Strauss G, Gibson W, Allison GT , McKay J

WFUMB Congress 2011 (World Federation for Ultrasound in Medicine and Biology), 26. - 29.08.2011, platform presentation; Vienna, Austria.
Title: 'Can M-mode ultrasound imaging of muscle tissue motion replace invasive electromyography for measuring muscle activation?' A. Dieterich, C. Pickard, L. Deshon, G. Strauss, W. Gibson, J. McKay.

Deutsches Symposium Forschung in der Physiotherapie, 26.11.2011, University of Göttingen, Germany.
Title: ‚Aktivierungsbeginn in den Hüftabduktoren: Messung mit M-mode Ultraschall' (Onset of hip abductor activation measured by M–mode ultrasound) Dieterich A, Pickard C, Deshon L, Strauss G, Gibson W, Allison GT, McKay J

International Summer University Osnabrück, 31.07.2012, Hochschule Osnabrück, Germany
Title: 'Gluteus minimus moves earlier in subjects with hip pain - a non-invasive study by M-mode ultrasound' A. Dieterich, C. Pickard, L. Deshon, G. Strauss, J. McKay.

IFOMPT Congress 2012 (International Federation of Orthopaedic Manipulative Physical Therapists), 30.9.-05.10.2012, platform presentation; Quebec, Canada
Title: 'Gluteus minimus activation is earlier in subjects with hip pain – a non-invasive study by M-mode ultrasound imaging' A. Dieterich, C. Pickard, L. Deshon, G. Strauss, J. McKay.

ZVK Stiftung zur Förderung von Forschung und Evaluation in der Physiotherapie 2012
10,000 Euro research grant for the project: ‚Entwicklung spezifischer Physiotherapie zur Behandlung der Koxarthrose: ein nichtinvasives Assessment zur Bewertung tiefer Muskelaktivität in Abduktorenübungen' (Developing specific physiotherapy for hip osteoarthritis: a non-invasive assessment of deep muscle activity in hip exercises)

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Abbreviations

ANOVA, analysis of variance

ASIS, anterior, superior iliac spine

B-mode, Brightness-mode

dGmed, deep gluteus medius

ECG, electrocardiography

EMG, electromyography

FFT, Fast Fourier Transform

Gmed, gluteus medius

Gmin, gluteus minimus

ICC, intra class correlation coefficient

M-mode, Motion-mode

MDC, minimal detectable change

MVIC, maximal voluntary isometric contraction

n, number

PSIS, posterior, superior iliac spine

r, Pearson correlation coefficient

r^2, determination coefficient

RMS, root of mean squares

SD, standard deviation

SEE, standard error of the estimate

VCR, video camera recording

US, ultrasound

1 Chapter One: Introduction

Overview of Chapter One

- Section 1.1 presents the clinical problem behind this research, hip pain.
- Section 1.2 introduces the relevance of the active control of hip joint mechanics and pinpoints the dearth of literature on the muscular control of hip joint mechanics and on the effects of hip pain on muscle activity. This section provides the rationale behind this research.
- Section 1.3 explains the selection of the gluteus medius and minimus muscles for this research.
- Section 1.4 states the research hypothesis.
- Section 1.5 introduces the research tools for assessing patterns of muscle activity.
- Section 1.6 states the aim of this research.
- Section 1.7 explains the definition and use of terms related to muscle activity.
- Section 1.8 presents the structure of the thesis.

1.1 Pain at the hip joint

The hip joint is a junction between locomotion and posture, uniquely shaped to allow movement in three degrees of freedom under high loading. Stabilization of the pelvis and trunk and simultaneous movement of the lower limb imply a complex muscular coordination. Hip pain limits the functional performance of the hip joint, constrains walking and active participation in social activities. Hip pain is a common problem with increasing relevance in ageing societies. The prevalence of hip pain in older Caucasian populations has been quantified between 11.9% (Cecchi et al 2008) and 19.2% (Dawson et al 2004). Hip pain is not confined to old age with 6.4% of an adolescent population sample reported hip pain (Spahn et al 2005).

Hip pain is a symptom that accompanies a range of pathologies (Bierma-Zeinstra et al 2001). Different regions around the hip joint may be symptomatic (Plante et al 2011, Tibor and Sekiya 2008). The main areas of hip pain are anterior, at the groin and lateral, around the greater trochanter (Birrell et al 2005, Bierma-Zeinstra et al 2001). Anterior hip pain is typically associated with the hip joint itself (Plante et al 2011). Hip pain in the groin area is a cardinal symptom of femoroacetabular impingement (FAI) (Sink et al 2008, Ganz et al 2003) and of pathology of the acetabular labrum (Burnett et al 2006). Both pathologies are considered predecessors of hip joint degeneration (Freehill and Safran 2011, Wagner et al 2003, Tschauner et al 2001). Chronic, non-traumatic, anterior hip pain in adults is indicative of osteoarthritis (Bierma-Zeinstra et al 2001) or a high risk to develop it. Physiotherapy is challenged to achieve pain reduction and prevent joint degeneration as long as possible, aims that require an in-depth knowledge of the dysfunctional mechanisms associated with anterior hip pain.

1.2 A perspective on the active control of hip joint mechanics

The recent description of the femoroacetabular impingement syndrome (Nogier et al 2010, Ganz et al 2003) and the technical progress in hip arthroscopy (McCarthy and Lee 2011, Tibor and Sekiya 2008) have increased the recent literature pertaining to diagnosis and surgical repair of structural lesions or deformities at the hip joint. Structural deformities that are known to increase the risk for hip pathology are hip

dysplasia (Jacobsen and Sonne-Holm 2005, Reijman et al 2005), acetabular retroversion (Ezoe et al 2006, Reynolds et al 1999) and an increased alpha angle (Johnston et al 2008). Large samples sizes are needed to demonstrate a significant pathogenic effect of theses deformities. Epidemiologic studies indicate that many individuals who bear the risk factors manage well and stay asymptomatic (Gosvig et al 2008, Jacobsen et al 2005, Lane et al 1997). The method to manage successfully a structurally demanding hip joint is of great interest for physiotherapy.

Current literature on the treatment of hip pain focuses on surgery to correct structural aberrations (Haene et al 2007, Sampson 2005, Ganz et al 2003). Recent study results question the assumption that structural corrections alone restore asymptomatic hip joint function (Luder et al 2011). However, the surgical perspective neglects that active hip joint control may be necessary to avoid cartilage shear and local overloading. The incongruent hip joint (von Eisenhart-Rothe et al 1999, Menschik 1997) provides joint play in order to enable a combination of gliding and rolling motions (Kubein-Meesenburg et al 1993). Hip joint play has been measured as the posterior-anterior glide of the femoral head in the acetabulum (Harding et al 2003) and as the displacement of the femoral centre of rotation with hip flexion (Gilles et al 2009). Few publications pertain to deficits in muscular control associated with hip pain (Kennedy et al 2009, Austin et al 2008, Zimny 1998). These case reports of the successful physiotherapy treatment of hip pain challenge the surgical treatment agenda (Austin et al 2008, Zimny 1998) and indicate the need for further research to specify muscular deficits in hip joint control.

Multiple layers of muscles surround the hip joint. Hip muscles are organized in synergistic groups of superficial and deep muscles with a similar line of pull. Study results suggest a redistribution of muscular activity between the superficial and deep hip abductors in hip joint degeneration (Grimaldi 2011, Sims et al 2002). A pain induced redistribution of muscle activity among synergists is endorsed by current theory on motor adaptations in painful conditions (Hodges and Tucker 2011, Arendt Nielsen and Falla 2009). Pain has an immediate and often lasting effect on muscle recruitment (Henriksen et al 2011, Le Pera et al 2001, Hides et al 1994) that aims to protect the painful area. Altered muscle recruitment may adversely affect joint stabilization and local joint loading in the long-term (Falla 2004). An understanding of the motor patterns associated with hip pain would provide a basis for developing targeted

physiotherapeutic interventions to improve active hip joint control and reduce local overloading.

1.3 The gluteus medius and minimus muscles

The limited evidence which has been published on pain induced alterations in active hip joint control refers to the hip abductors (Grimaldi et al 2009b, Sims et al 2002, Fredericson et al 2000). During walking, the hip abductors are challenged to stabilize up to 2.5 times body weight in regular loading (Bergmann et al 2001). In the gait phase of 'loading response' (Perry 1992) this high load needs to be balanced on the femoral head of the stance leg whilst bringing forward the trunk and the free leg. Hip joint stabilization in the frontal plane is mainly achieved by the gluteus medius and minimus muscles, the prime hip abductors and pelvic stabilizers. This pair of synergistic muscles, similar in topography and line of pull, is organized in a superficial and a deep level. The functional differences of gluteus medius and minimus need further investigation to understand their individual roles in hip joint control.

1.4 Research hypothesis

The main hypothesis of this research is that subjects with non-traumatic, anterior hip pain activate the gluteus medius and minimus muscles in a different motor pattern than asymptomatic controls.

1.5 Tools for assessing patterns of muscle activity

The current gold standard for measuring muscle activity is electromyography (EMG). Whilst the application of EMG is relatively uncomplicated on superficially accessible muscles, invasive needle or fine-wire EMG must be used for deep muscles. The insertion of fine-wire electrodes is an invasive and cumbersome procedure that includes the risks of vessel or nerve injury, infection and fine-wire breakage (Hodges 2006, p. 21) and is for ethical reasons limited to the application by experts on small samples mainly for research purposes. Fine-wire EMG is not an applicable assessment

tool in rehabilitation. It cannot be used to assess individual activation patterns, to evaluate a patient's response to different exercises and to document whether a therapeutic intervention achieved a change of activity. An assessment method is needed that enables monitoring activity patterns within physiotherapy treatment. Further, an ethically less problematic method is required for examining motor patterns in clinically relevant sample sizes. Ultrasound (US) imaging is used successfully in several body regions to assess deep muscle activity non-invasively (Whittaker 2007, Thompson et al 2006), but not at the hip joint.

1.6 Aim of this research

This research aims to develop an US assessment with clinical applicability for determining activity patterns of the gluteus medius and minimus muscles.

1.7 The structure of the thesis

This introductory first chapter is followed by a two-part literature review. The first part reviews methods to measure muscle activity with a focus on US imaging. A second part presents the functional anatomy of the gluteus medius and minimus muscles.

The research process was structured by a sequence of questions. The outcomes of these questions informed each stage of the research (Figure 1.1). Chapter Three reports the development of the methods used in the two experimental setups in which the data for the studies on US methodology were collected.

The most established US measurement in rehabilitation research is muscle thickness. Muscle thickness has been measured using B (brightness)-mode and M (motion)-mode US. It was unclear which US mode is advantageous for measuring the change of muscle thickness associated to muscle activity. Chapter Four presents Studies One and Two which relate to the choice of the US mode and the scanning plane for measuring gluteus medius and minimus thickness.

The determination of the activation level from US measurements of thickness change requires the quantification of prediction. Chapter Five reports Study Three, the prediction of the activation level of gluteus medius and minimus, as measured by dynamometry, surface and fine-wire EMG, from the percentage of thickness change measured using M-mode US. The research results on thickness measurements are summarized in Chapter Five.

The literature reports measurements of the onset of muscle activity using M-mode and Doppler US. It is unclear whether M-mode or Doppler US, which are based on different processing techniques of the US echoes, is more sensitive to detect muscle motion. Chapter Six reports Study Four in which reliability and method agreement of M-mode, Pulsed-Wave Doppler and surface EMG measurements of the onset of gluteus medius activity were investigated.

The M-mode image of the activation process indicates different M-mode greyscale patterns. To overcome limitations of the US measurements of motion onset, innovative measurement variables were developed. These variables are based on the visually detectable greyscale patterns in the M-mode traces of muscle activity. Study Five in Chapter Seven investigated the hypothesis that M-mode high-frequency patterns are indicative of electrical excitation of muscle. A computed detection algorithm for high-frequency patterns was developed and the relationship between high-frequency patterns and activation measured by fine-wire EMG was examined.

Finally and to come back to the clinical question behind this research, M-mode high-frequency patterns were applied in Study Six, a case-control study, to assess the activity of gluteus medius and minimus in subjects with and without anterior, non-traumatic hip pain. Study Six is reported in Chapter Eight.

Chapter Nine summarizes and discusses the research findings, outlines the limitations of this research and recommends possible directions for future research. The chapter ends with the final conclusions of the thesis.

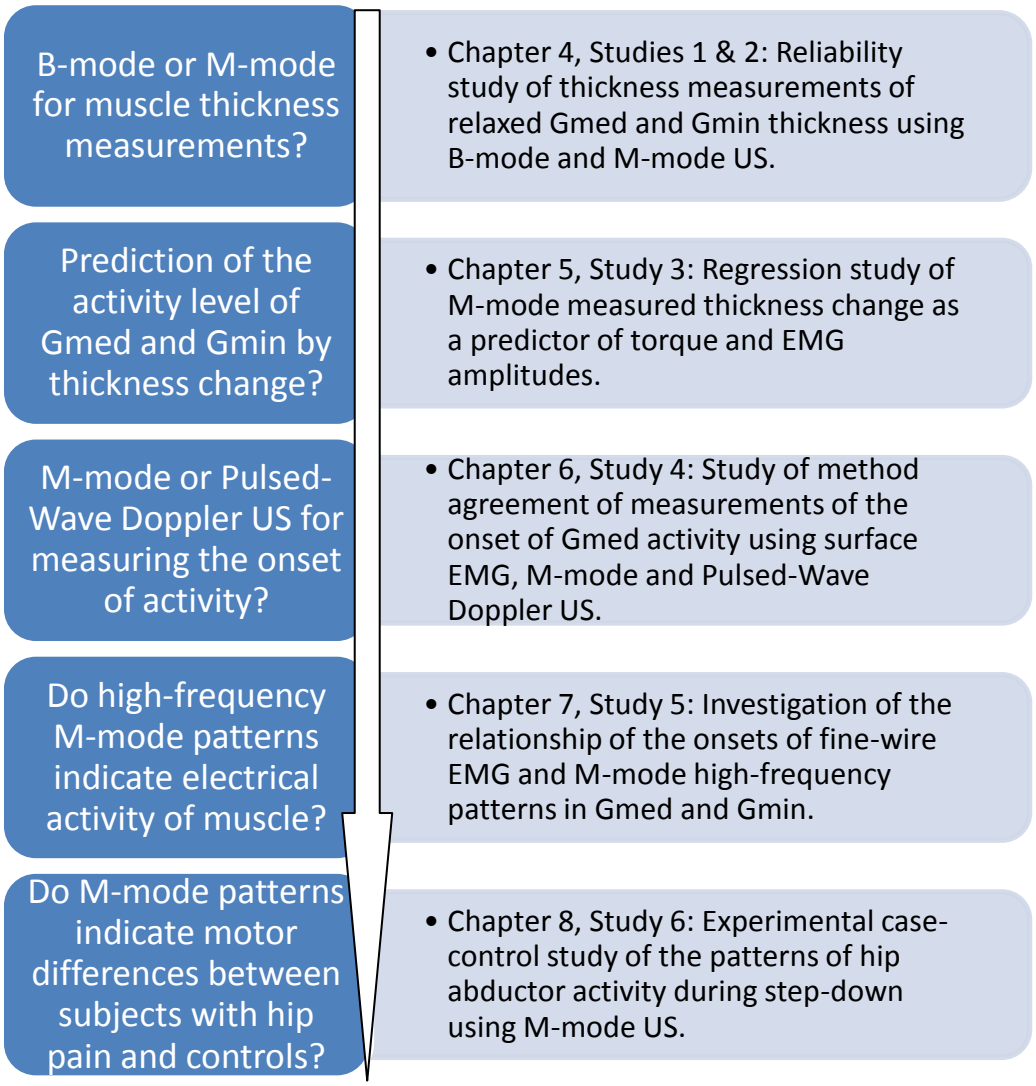


Figure 1.1. Guiding questions determined the structure of the experimental research. Abbreviations: B-mode, Brightness-mode; EMG, electromyography; Gmed, gluteus medius; Gmin, gluteus minimus; M-mode, Motion-mode; US, ultrasound.

1.8 Terms in the context of muscle activity

Muscle *activation* or *activity*, muscle *activation patterns* and *motor patterns* are commonly used terms. Their use in this thesis needs to be defined.

The concept of muscle activation refers to the physiological processes linked to force generation by muscles. These processes are based on the model of 'excitation-contraction coupling' (Lieber 2010, p. 41) and include a chain of electrical, chemical and mechanical events. At the muscle itself, activation starts with acetylcholine release in the synaptic cleft that triggers a depolarization of the muscle fibre sarcolemma and the propagation of action potentials along the muscle (Lieber 2010, pp. 41-43). Membrane depolarization results in the release of calcium ions which trigger the cross-bridge cycle, the central element of muscle contraction. With the drop of the calcium level the muscle relaxes and the activation finishes (Lieber 2010, pp. 43/44). In this thesis, the term *muscle activity* is used to express the complex activation process including electrical and mechanical phenomena.

Awareness that muscle activity includes electrical and mechanical aspects is important when comparing EMG and US measures of activity, because these modalities are sensitive to different aspects of muscle activity. EMG, the most common method to quantify muscle activity, measures changes in the electrical potential of muscle tissue that are associated to the activation process. Commonly, the term *muscle activation* is used synonymously with electrical excitation of muscle. Electrically measurable muscle activity is considered the proof of movement being *active*. In this thesis, the terms *electrical excitation* or *action potentials* are used to describe the electrical aspect of muscle activity that is measured by EMG.

US is sensitive to mechanical phenomena associated with muscle activity, such as the displacement and motion of muscle tissue. The term *muscle deformation* has been proposed for these mechanical phenomena (Vasseljen et al 2006). However, the term deformation implies that the deforming energy is not part of the object that is deformed. Otherwise, it is said to be a transformation. The source of the energy which initiates and sustains a change of muscle shape or muscle motion cannot be defined from an US image and may be within the muscle or from elsewhere and acting on the muscle. Therefore, the terms *muscle motion, thickness change* or *mechanical activity*

are used in this thesis to describe mechanical phenomena associated with muscle activity and detectable by US.

The term *muscle activation pattern* has been used with regard to different types of patterns. For example, an activation pattern may refer to the amplitude of electrical excitation of single muscles over time (Lacquaniti et al 2012), or to intra-muscular differences in timing and amplitude of electrical excitation (Watanabe et al 2012) or to differences in electrical excitation between synergistic (Day et al 2012) or antagonistic (Bazzucchi et al 2008) muscles. Most commonly, activation patterns are derived from EMG measurements. In this thesis and in accordance with the literature, the use of the term *activation pattern* refers to patterns of electrical excitation. The terms *motion pattern* or *activity pattern* are adopted for US derived patterns.

2 Chapter Two: Literature Review
Tools for Assessing Muscle Activity: Dynamometry, Electromyography and Ultrasound Imaging

Overview of Chapter Two

- Section 2.1 gives an introduction into the topic and the structure of the literature review.
- Section 2.2 presents a short overview of dynamometry which measures the force produced by muscle activity.
- Section 2.3 introduces electromyography, the current standard for measuring the level and the onset of the electrical excitation during muscle activation.
- Section 2.4 reviews measurements of structural and dynamic parameters of muscle by ultrasound imaging. Several measurements in different ultrasound modes have been applied, which are directly or indirectly linked to the performance of muscle. This section also discusses the specific risks of ultrasound applications when measuring muscle activity, and appropriate safety measures.
- Section 2.5 describes the gluteus medius and minimus muscles that are the target muscles of this research.
- Section 2.6 summarizes the main points of the review.

2.1 Introduction

The main hypothesis of this research is that subjects with anterior hip pain activate their gluteus medius and minimus muscles in an atypical intensity and timing. The investigation of this hypothesis requires the comparative assessment of superficial and deep hip abductor activity.

The intensity and the timing of muscle activity can be quantified by different methods, which were utilized in this research and which are reviewed here in the order of their historical appearance. First, dynamometry is introduced. Dynamometry is the objective quantification of the most obvious method to assess muscle activity, by gauging the effect of the effort, the resultant force. Second, electromyography (EMG) is reviewed, the gold standard with which selective muscle activation is measured by gauging the electrical excitation of muscle tissue. The current form of EMG has been developed over the past 60 years. The third and most extensively reviewed method to assess muscle activity is ultrasound (US) imaging. Static assessments of muscle diameter started approximately 40 years ago. In the past 20 years, dynamic applications have evolved. The application of US imaging is considered harmless. However, long-term recording of muscle activity increases the thermal risk. The risks of the application of US imaging on muscles and their safe management need consideration.

The finishing section of this review relates to the target muscles of this research and presents the functional anatomy of the gluteus medius and minimus muscles.

2.2 Dynamometry

Muscles activate to produce *force*. Force cannot be measured directly but is quantified by its effects, for instance the pressure against a dynamometer (Herzog 2007a, p. 311). Muscle activity produces a force that acts through the bone on which the muscle inserts, in a physics perspective, a lever. Muscle force results in a rotational movement or pressure of the respective limb. The outcome of muscle activity is defined as *torque*. Torque (τ) is the product of force (F) and the perpendicular lever length (r) upon which the muscle acts:

$$\tau = F\ r\ sin(\alpha),$$

Formula 2.1. Torque; sin(90°) = 1 (Rybach 2008, p. 47)

The practical relevance of this formula is that torque cannot be compared as an absolute value because it depends not only on force production but also on lever length. The measurement of lever length in the human body includes uncertainties, such as the exact position of the centre of rotation and the estimation of the bony endpoint. For simplification, torque measurements are commonly not compared directly, but are normalized as percentage of the individual, maximal, isometric torque production (maximal voluntary isometric contraction, MVIC).

A traditional technique to assess muscle function is the manual test of a muscle's capacity to produce torque (Montgomery and Hislop 1999). In order to gain a more objective quantification, pressure sensitive devices, dynamometers, are used. A dynamometer contains pressure sensitive elements which transform physical strain into electricity (Hennig and Lafortune 1997, p. 109). Sensors commonly used in rehabilitation are strain gauges or piezoelectric elements (Herzog 2007a, p. 318). Typical dynamometer devices in rehabilitation are force plates to measure ground reaction force and handheld dynamometers to measure torque. The output of dynamometric measurements is force amplitude over time which enables the quantification of the intensity and timing of torque production.

Some conditions and limitations apply to dynamometric measurements. To ensure the validity of the measurements, the dynamometer needs to be calibrated. In the calibration process, measured values are compared to known reference values (Deutsches Institut für Normung 1995a, part 1 p. 22). The measurement error, the

linearity of measurements and a point of intersection close to zero are controlled.
The range in which the measurements of a dynamometer are accurate is limited. Normative values (Andrews et al 1996) allow for appreciation whether the range of a device is sufficient for the intended test.
Most handheld dynamometers are sensitive only for the perpendicular dimension of an applied force, whereas floor integrated force plates, a specific subgroup of dynamometers, mostly display three-dimensional force components.
Dynamometers are pressure sensitive regardless of the source of pressure. The tester must ensure that the measured force results from muscle activity and not, for example, from the weight of the limb leaning against the pressure transducer.

Common and scientific language differs in the context of torque measurements. The measurement of ‘maximal voluntary isometric contraction’ does not measure the contraction but its effect. The effect of muscle contraction is commonly labelled ‘force’ or ‘strength’ although it is per physical definition ‘torque’. In contrast, the output of dynamometric measurements can generally be termed ‘force’, because the dynamometer cannot distinguish the source of pressure. Pressure can originate from torque or from body, limb or object weight.

Dynamometers measure the composite result of synergistic activity, the contribution of single muscles cannot be distinguished. To measure selective muscle activity, electromyography is the current gold standard.

2.3 Electromyography

2.3.1 Introduction

The established method of assessing single muscle activation is electromyography (EMG). According to Basmajian and De Luca, the first statement on muscle-generated electricity dates back to the Italian Redi in 1666 and referred to the electric ray fish. The first experimental proof of the electric nature of muscle contraction was performed by Galvani in 1791 (Basmajian and De Luca 1985, p. 1). The advent of modern EMG methods to explore in-vivo muscle activation was established with the textbook 'Muscles alive', first edited in 1962 (Basmajian and De Luca 1985).

EMG records electrical potentials in reach of the electrode(s). There are two main types of EMG, the non-invasive surface EMG, in which electrodes are attached to the skin overlying the muscle of interest, and the invasive EMG by fine-wire or needle electrodes, which are positioned in the muscle tissue itself. This review does not refer to needle electrodes which are rarely used in research for rehabilitation.

EMG in its current form comprises the development of elaborate techniques to probe and process raw, electrical signals towards accessible information on muscle activity.

2.3.2 The technical process to separate the electromyographic signal of muscle excitation

The electrical signals received by the EMG electrodes contain not only the very small electrical potential of muscle excitation but also signals from the environment and the equipment which are summarized under the term *noise* (Basmajian and De Luca 1985, p. 37). The quality of an EMG signal is determined by the *signal to noise ratio* that expresses the amplitude of the signal of muscle excitation as a multiple of the baseline signal during relaxation (Basmajian and De Luca 1985, p. 52). Over the past 60 years, a sophisticated technique and suitable equipment have been developed to separate, amplify and process the muscle signals.

The relative separation of the signal that originates in muscle activity is achieved in several steps starting with the differential recording technique. Two electrodes on or in the target muscle receive two slightly time-shifted and therefore

different signals of muscle excitation plus the same signal of noise from the environment and equipment. The part of the signal that is common to both electrodes, the noise, is subtracted, a process termed *common mode rejection* (Cram and Kasman 1998, pp. 49/50). The remaining signal is the difference of the two signals and is related primarily to muscle excitation. This remaining signal is amplified. By the differential recording technique, noise is reduced, but not eliminated (Basmajian and De Luca 1985, p. 53). Noise cannot be separated fully from the muscle signal.

The next step is filtering the signal. Filters reduce the range of frequencies contained by the signal. A basic rule in physics is that the valid representation of frequencies depends on the sampling rate. According to the Nyquist/Shannon theorem, only frequencies in up to half of the sampling rate are represented unbiased, whereas aliasing artefacts occur in the upper half of the spectrum (Merletti et al 1999a). At least the upper half of the spectrum of the EMG signal is therefore cut-off by a low-pass filter. Filtering recommendations differ for surface and fine-wire EMG because the frequency content of the signals differs. The surface EMG signal passes through muscle, subcutaneous tissue and skin before it is received by the electrodes. These tissues act as low-pass filters, which is the reason that a fine-wire EMG signal contains much higher frequencies than a surface EMG signal (Winter 1990, p. 199). A surface EMG signal contains in 95% frequencies up to 400 Hz (Merletti et al 1999b). Low-pass filtering between 500 and 1000 Hz, depending on the sampling rate, has been recommended (Merletti et al 1999a). To replicate its content of higher frequencies, fine-wire EMG must be sampled at a higher frequency and filtered at a higher cut-off point. A sampling frequency of at least 2000 Hz and a cut-off at 1000 Hz has been advised (Hodges 2006). The lowest frequencies of the EMG signal contain *movement artefacts*. These unwanted signals may be removed by high-pass filters (Cram and Kasman 1998, pp. 50/51). A cut-off between 10 and 20 Hz has been recommended (Hodges 2006, Merletti et al 1999a). Additional band-stop filtering may be advisable if a frequency analysis indicates peaks of electrical noise within the frequency range of the muscle signal (Figure 2.1).

After these cleaning processes, the remaining EMG signal is considered the best achievable reflection of motor unit action potentials in the recording area of the target muscle (Cram and Kasman 1998, pp. 49/50). Further processing steps are optional and depend on the aim of the EMG analysis.

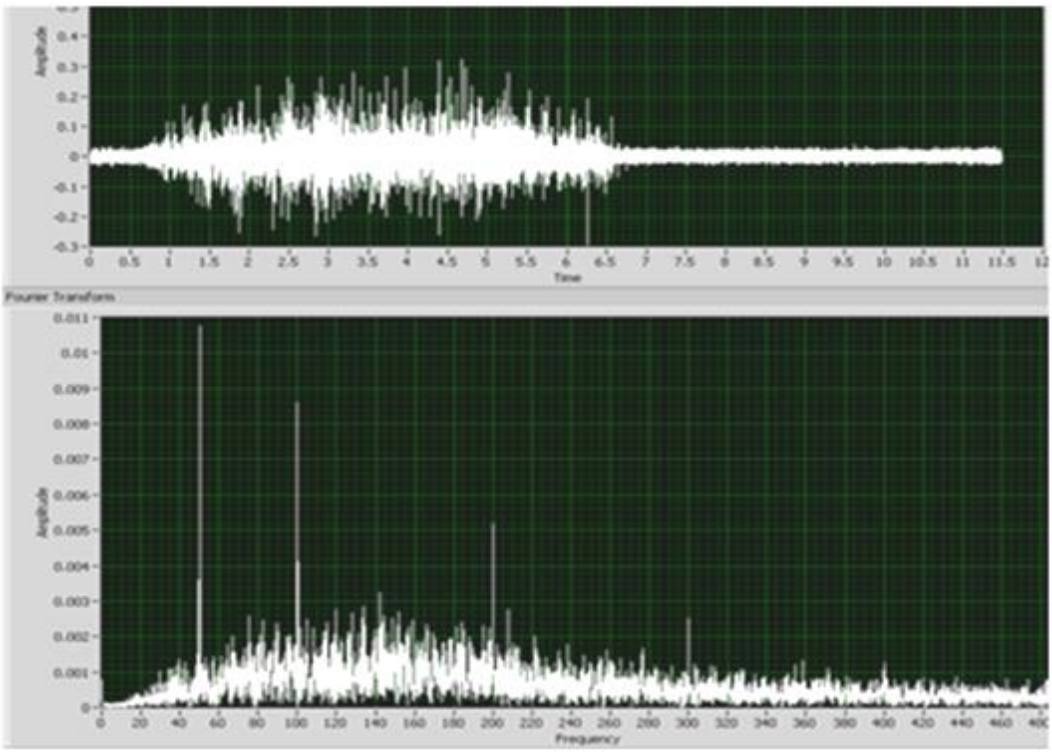


Figure 2.1. Frequency analysis (bottom window) of the surface EMG signal in the top window indicates contamination with noise at 50 HZ and the harmonics of 100 Hz. The signal can be 'cleaned' using narrow band-stop filters.

2.3.3 EMG analysis

The EMG signal contains information regarding the beginning and the end of excitation, the neural drive to the muscle, the level of effort to produce torque, the ability to relax and also the fatigue of a muscle (Hodges 2006). EMG signals can be analysed in the *frequency* and in the *time domain*. Frequency analysis is primarily performed to detect muscle fatigue (De Luca 1997, Mortimer et al 1970) and is not a topic of this review.

In the time domain, EMG signal amplitude is displayed over time providing an indication of the *intensity and the timing of motor unit action potentials* in the recording area (Soderberg and Knutson 2000). The intensity of electrically measurable activation can be quantified by different procedures. The choice depends on the type of recorded activation.

Broadly applicable is the *linear envelope*, which has been defined as the standard for comparing EMG to force or kinematic data (Merletti et al 1999b, Winter

1990). The linear envelope is a strongly smoothed EMG signal which results from full-wave rectification and low-pass filtering with a cut-off below 10 Hz (Winter 1990, pp. 204/5). The resulting signal resembles a force trace and is closely related to it, in spite of some limitations in the EMG amplitude – force relationship (De Luca 1997). Arguments against the use of linear envelope processing are that most signal information is filtered away and that the temporal accuracy is reduced by the loss of information (Merletti et al 1999a).

In static activity, e.g. sustained, isometric activations, extraction of the level of excitation by calculation of the *root of mean squares (RMS)* has been recommended (Hodges et al 2003, Cram and Kasman 1998, pp. 59/60, Basmajian and De Luca 1985, p. 97). The RMS calculation results in a single amplitude value for a predefined window of time. The amplitudes of the data points in the time window are squared, the mean of all squares is calculated and then the square-root is taken,

$$RMS = \sqrt{\frac{x_1^2 + x_2^2 + \cdots + x_n^2}{n}}$$

Formula 2.2. The root of mean squares (RMS); n, number of the values from x_1 to x_n which form the window of analysis.

EMG signal amplitude depends not only on the level of excitation but also on the measurement conditions, for example the thickness of the subcutaneous tissue, the direction of the bipolar electrode arrangement and the location of the electrodes on the muscle (De Luca 1997, Solomonow et al 1986). Movement of the target muscle influences the stability of the EMG signal and signal amplitude. Inferences from EMG amplitude measured during anisometric muscle contractions are therefore limited (De Luca 1997). Due to this dependency of EMG amplitude on the measurement conditions, a *normalization* procedure to a standard intensity of activation is generally performed when EMG and torque amplitude are compared (Bolgla and Uhl 2007, Burnett et al 2007).

Temporal parameters of activation, such as the *onset and duration of excitation*, can be detected from the raw or rectified signals, without further processing (Tsao et al 2009, Hodges 2006). To detect the onset of activation, a sufficiently long baseline signal of the relaxed muscle is needed. Commonly, the onset is defined as the first rise of the amplitude over the baseline or over a baseline derived threshold value (Allison 2003,

Hodges and Bang 1996). This concept of the onset of activation is theoretically clear, but under the conditions of real-world performance often difficult to apply. Incomplete relaxation, heightened baseline activity, small pre-emptive bursts of electrical excitation or artefacts from cardiac activity cause ambiguous or unclear onsets (Lee et al 2007a, Hodges 2006, Vasseljen et al 2006, Allison 2003). Complex computer algorithms have been suggested to overcome these difficulties (Conforto et al 2006, Staude and Wolf 1999, Bonato et al 1998) but they are complicated to apply. Due to difficulties to detect valid, automated onsets of activation, it is common practice to detect the onset of excitation visually (Vasseljen et al 2006, Falla et al 2004b) or to confirm computer detected onsets visually (Bolgla et al 2010).

2.3.4 Surface EMG versus fine-wire EMG

Surface EMG and fine-wire EMG signals of the same activity are not equivalent (Rudroff et al 2008, Rechtien et al 1999). The signal content differs. A surface EMG signal is a global representation of the electrical excitation in reach of the electrode. The exact source of the measured excitation is undetermined; it may originate in the superficial, in deeper or even in adjacent muscle and the source of the signal may have changed by muscle movement (De Luca 1997).
A fine-wire EMG signal represents local motor unit action potentials of a few cubic millimetres of muscle into which the hooked ends of the wires are fixed. Due to regional differences in activation, the fine-wire signal may not be representative for the muscle in total (Hug 2011). However for local excitation, the fine-wire EMG is a more complete representation than achievable in surface EMG. The fine-wire EMG signal contains high frequencies, which have been filtered during their passage through muscle and subcutaneous tissue in surface EMG. Fine-wire EMG is less affected by ‘noise’, as signal reception occurs in the muscle itself.

The EMG procedure and the associated inconvenience differ. Surface EMG recording, in particular dynamic recording of several muscles, requires the application of many electrodes. The use of cables may affect typical motor behaviour. The insertion of invasive fine-wire EMG may be painful and the small injury by fine-wire insertion may induce a protective behaviour.
Young et al (1989) investigated changes of gait parameters by the application of surface and fine-wire electrodes in 38 children with spastic diplegia. A significant decrease in cadence was found after application of surface EMG, followed by further significant

decreases in step length and walking velocity after fine-wire insertion. Other studies which compared activation patterns recorded by surface and fine-wire electrodes led to conflicting results. Jacobson et al (1995) found no significant differences in ensemble-averaged linear envelopes of surface and fine-wire EMG patterns of quadriceps and biceps femoris during walking and jogging. Bogey et al (2000) reported significantly different timing patterns by the two electrode types but no significant differences for activation intensities of the tibialis anterior and soleus muscles. Chimera et al (2009) stated no significant onset differences but significant differences in the intensity of activation of the tibialis anterior, gastrocnemius and soleus muscles. In conclusion, there are trends for differences between surface EMG and fine-wire EMG signals of the same activity and these differences may be significant.

2.3.5 Limitations of EMG

Principally, EMG is limited to measure the electrical aspect of muscle activity. In complex tasks, evidence supports an effect of both types of EMG on the object of measurement, motor behaviour (Young et al 1989). The most relevant limitation of surface EMG is the limited accessibility of muscles. According to Cram and Kasman, only 68 of more than 600 human muscles can be accessed by surface EMG (Cram and Kasman 1998, p. 221).

The most relevant limitations of fine-wire EMG arise from the invasive procedure and the associated risks of damage to nerves and vessels, infection and pain. Further risks are vasovagal episodes, e.g. fainting, and the breakage of the fine-wire electrode, with the consequence that the distal part remains within the muscle (Hodges 2006, Basmajian and De Luca 1985, p. 32). These risks constitute ethical limitations in the application of fine-wire EMG and are important reasons to invest efforts into the development of non-invasive alternatives.

The next section of this review presents the development of musculoskeletal ultrasound imaging towards the non-invasive assessment of deep muscle activity.

2.4 Musculoskeletal Ultrasound Imaging

2.4.1 Historical development

The use of ultrasound (US) imaging for medical purposes dates back to the late Thirties of the last century when Karl Dussik, an Austrian psychiatrist, developed ultrasonic measurements of an appearance within the skull, which he interpreted to be the ventricles (Dussik 1952, Dussik 1942). Learning from errors and about ultrasonic artefacts, the imaging method, which he termed 'hyperphonography', developed. The first musculoskeletal application of US imaging was a report on the measurements of articular tissues (Dussik et al 1958).

The further development of imaging by US, which established as 'sonography', is intimately related to the technical improvements of US imaging equipment and the deepened understanding of the physics behind the imaging process. This development allowed for progress from time-extensive scanning in a water bath in order to gain a rough, static, black and white image, towards direct contact, real-time scanning resulting in well defined, greyscale tissue sections (Thomas et al 2005). Some landmarks during this process were the definition of thickness measurements and tissue density (Wild 1950), real-time processing (Krause and Soldner 1967) and the greyscale technique (Kossoff and Garrett 1972). A milestone for the clinical acceptance of US imaging (Thomas et al 2005) occurred when Donald was able to diagnose the inoperable stomach cancer of a patient to be an ovarian cyst (Donald et al 1958).

In the Eighties, the development of equipment and knowledge allowed for a broader clinical use of sonography (Frentzel-Beyme 2005). The screening of newborns for congenital hip disease was an early musculoskeletal US application of epidemiologic importance (Walter et al 1992, Graf 1980). In 1991, the first comprehensive textbook on musculoskeletal US was published (Van Holsbeeck and Introcaso 2001). Since then the number of applications has expanded. Specific musculoskeletal US applications beside two-dimensional B-mode imaging include the use of Colour Doppler sonography to detect inflammatory processes, US guided injection techniques, ultrasonic contrast agents and Elastography for distinguishing mechanic tissue properties (Kane et al 2004). New technologies and applications are still emerging.

2.4.2 Focus

As reflected in US textbooks, the current focus of musculoskeletal US imaging is the medical diagnosis of irregularities, lesions or disorders by a structural assessment, sometimes supported by imaging of passive or active movement (Bianchi and Martinoli 2007, Van Holsbeeck and Introcaso 2001). The professional assessment of muscle activity is not part of medical sonography education but a specific interest of physiotherapists (Whittaker 2007) and other professions working on motor activity.

2.4.3 Principals of ultrasound imaging of skeletal muscles

US imaging of skeletal muscles commences with anatomic orientation using the two-dimensional B-mode. Textbooks describe standards according to patient and transducer position to achieve an indicative image. Other US modes, in particular Colour Doppler to visualize vascularisation, may be added for additional information. Muscles should be assessed during relaxation and activation using the longitudinal and the transverse scanning planes (Zamorani and Valle 2007, p. 47).

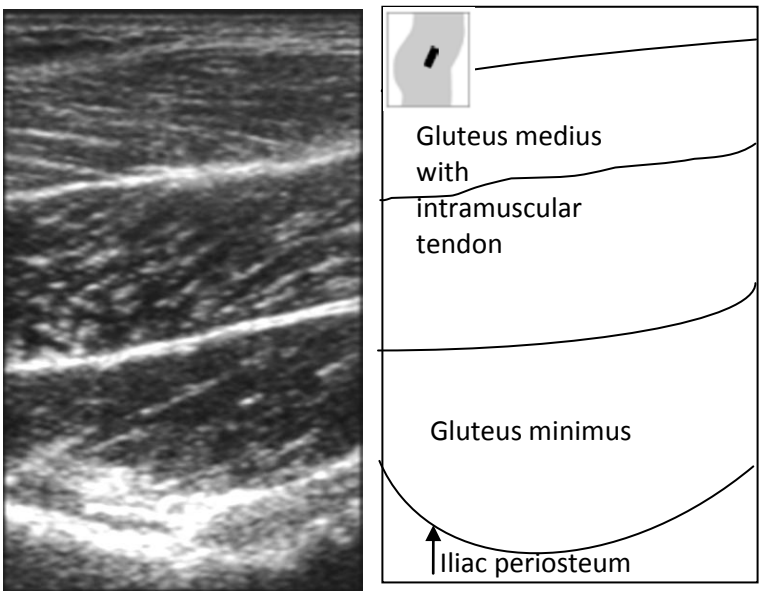


Figure 2.2. B-mode image, longitudinal scanning plane, section of gluteus medius and gluteus minimus. Note the fascicular structure of the muscle tissue.

The B-mode US image is a delineation of interfaces between tissues with different sound impedance (Hoskins et al 2010, Stokes et al 1997). Bright interfaces are referred to as hyperechoic, indicating their high reflective capacities, whereas dark, non-reflective tissue is termed hypoechoic. The image of a muscle is characterized by the hyperechoic tendon and muscle fascia, the hyperechoic, fibrous, long, thin lines of

fascicles and the hypoechoic contractile tissue (Walker et al 2004) (Figure 2.2). The reception of the reflected sound pulses, the echoes, is best when the probe is orientated perpendicular to the structures of interest. In highly reflective tissues, such as muscle and tendon, a slight tilt of the scanning angle causes a marked, nonlinear change of the greyscale delineation in the US image. This phenomenon is termed anisotrophy (Van Holsbeeck and Introcaso 2001, pp. 17/18). Anisotrophy can be used to accentuate a muscle's image by increasing the contrasts (Zamorani and Valle 2007, p. 48).

Musculoskeletal US imaging is the starting point for assessing muscle function, as it describes and defines standard views and approaches to the imaging of muscles by US. Structure and function of a muscle are inextricably related. Assessment of the activity of a muscle without observing its structure is not possible. The professional use and responsible interpretation of sonography is bound to legally stipulated conditions. Physiotherapists who use US imaging without having acquired a sonography qualification must act and interpret cautiously and in awareness of their professional boundaries. [1] The process of formalizing physiotherapists' use of US imaging is ongoing.

2.4.4 Rehabilitative ultrasound imaging

An international symposium in 2006 termed the use of sonography by physiotherapists *Rehabilitative Ultrasound Imaging*, abbreviated RUSI. RUSI was defined as

"a procedure used by physical therapists to evaluate muscle and related soft tissue morphology and function during exercise and physical tasks. RUSI is used to assist in the application of therapeutic interventions aimed at improving neuromuscular function. These purposes include the use of US for providing feedback to the patient and physical therapist to improve clinical outcomes. Additionally, RUSI is used in basic, applied and

[1] US based diagnosis of musculoskeletal pathology is the legal responsibility of medical doctors or allied health professionals who fulfil conditions regarding training and experience regulated by national or supra-national organizations, e.g. the Australasian Sonographer Accreditation Registry (ASAR), the British Medical Ultrasound Society (BMUS) or the American Registry for Diagnostic Medical Sonography (ARDMS). The incidental observation of structural irregularities by physiotherapists may provoke a competitive situation which must be handled in the patient's interest and within professional boundaries. In this sense, Whittaker published via internet a consensus form for physiotherapists who apply US imaging.
http://rtuspt.com/documents/RUSIConsentForm.pdf. (Whittaker, J. L.; retrieved 2011/02/15)

clinical rehabilitative research to inform clinical practice." (Teyhen 2006).
The assessment of muscle activity to inform physiotherapeutic practice is within the scope of RUSI.

It is important to note that RUSI is distinguished not by a specific, unique application but by its purpose. Some professions beside physiotherapists investigate muscle activity by US, for example biomechanists and physiologists (Mannion et al 2008a, Reeves and Narici 2003, Kubo et al 2000, Fukunaga et al 1997) or sports scientists (Karamanidis et al 2005).

2.5 The assessment of muscle activity by ultrasound imaging

2.5.1 Trends in ultrasound imaging of muscle activity

In July 2012, a literature search was undertaken to examine basic trends in the published literature on US imaging of muscle activity. This simple search in Pubmed with the keywords "*ultrasound imaging" muscle (activ* OR onset OR contraction)* retrieved 120 publications related to US imaging of skeletal muscle activity (Appendix 2.1). The retrieved publications reflect basic developments in US imaging of muscle activity. These developments include a marked rise in publications over the past ten years (Figure 2.3), a focus on trunk muscles (47% of publications) (Figure 2.4) and the primary use of thickness measurements as the assessment parameter (66% of publications) (Figure 2.5).

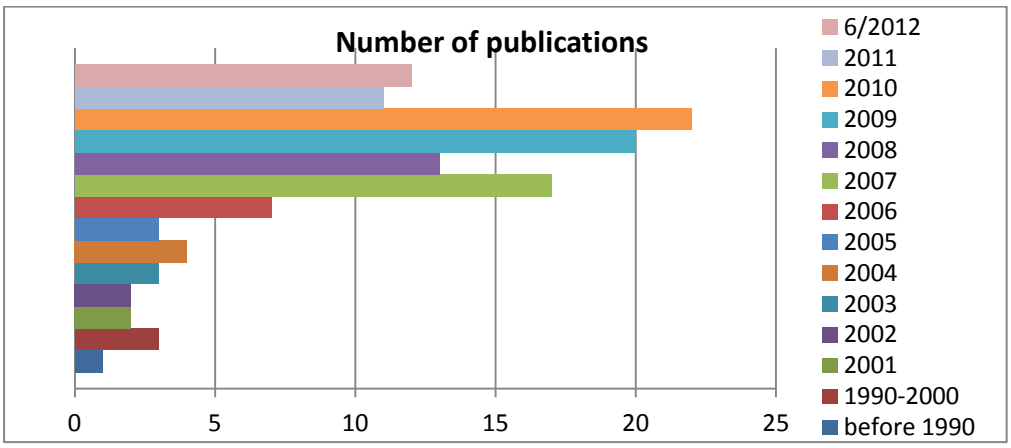


Figure 2.3. Number of publications related to US imaging of muscle activity per indicated time period.

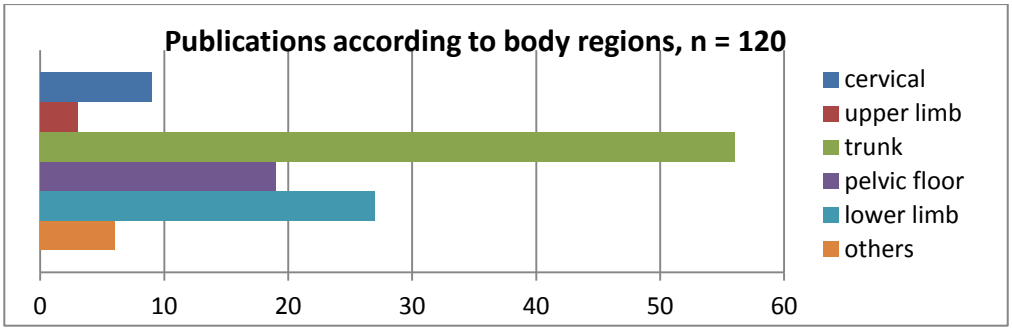


Figure 2.4. Number of publications related to ultrasound imaging of muscle activity according to body regions.

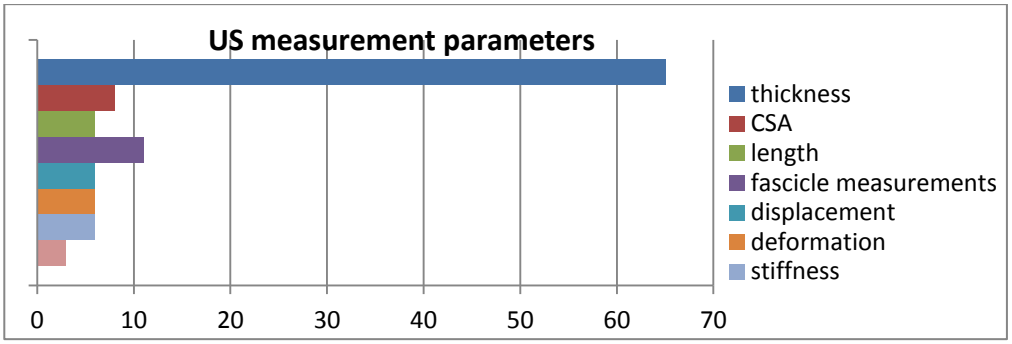


Figure 2.5. Number of publications in which one or more of the listed US parameters to measure muscle activity has been used, publications on pelvic floor muscles not included (n=98, multiple parameters possible).

2.5.2 Approaches to assess muscle activity by ultrasound imaging

The US image is a greyscale translation of sound pulses. An US imaging system records the time when a sound pulse is sent, the time when an echo is received, the amplitude, the waveform and the spectrum of the received echo. From these parameters, calculations of the position, the echogenicity and the motion of an interface, such as muscle fascia or perimysium, can be calculated. In activating muscle, the contraction of the sarcomeres induces movement and displacement of interfaces and also changes in shape of muscle. These effects of muscle contraction are measurable using US imaging. Different measuring approaches have been developed which are introduced in the following sections according to their historical development. Static measurements of muscle are included, because they are considered indirect indices of muscle use (Stokes and Young 1986), and they build the basis on which dynamic measurements developed.

The following review is divided into two sections, one related to measurements of structural parameters that change by muscle activity and a second related to measurements of muscle motion.

2.5.3 Cross-sectional area

Measurements of the cross-sectional area (CSA) exist in two types, the anatomical and the physiological CSA. Both the anatomical and the physiological CSA are related to muscle volume, which is a major determinant of the maximal torque that can be produced by a muscle (Fukunaga et al 2001). The anatomical CSA is a cross-section perpendicular to the long axis of the muscle and can be imaged using US, computed tomography (CT) or magnetic resonance imaging (MRI). The latter is considered the gold standard (Esformes and Narici 2002, Hides and Richardson 1995). CSA measurements by US have been proven to correlate highly with those taken by MRI (Dupont et al 2001, Hides and Richardson 1995). If not specified otherwise, CSA measurements refer to the anatomical CSA of a muscle. The physiological CSA is a cross-section perpendicular to the long axis of the muscle fibres and must be calculated on the basis of the architectural parameters of a muscle (Formula 2.3). The physiological CSA is considered the morphological correlate of a muscle's capacity to generate torque (Lieber 2010, p.27). The physiological CSA is in relation to muscle strength more relevant than the anatomical CSA, but it is not a 'ready to use' parameter and the calculation variables are often unknown.

Physical strength is defined as the capacity of muscles to generate torque (Lieber 2010, pp. 103-6). Numerous studies confirmed a positive, albeit limited, correlation between a muscle's maximal torque and its anatomical CSA. For example, Sale et al (1987) reported correlation coefficients between 0.58 and 0.8 for the biceps brachii, and Miller et al (1993) detected a correlation of 0.84 for the quadriceps and of 0.95 for the elbow flexors. Young et al (1984) estimated a correlation between 0.53 and 0.66 for the CSA of female quadriceps muscles and 0.77 for the quadriceps of older men, but only 0.15 for young men's quadriceps (1985).

Limitations of the moderate to good correlation between strength and CSA were indicated by Young's finding of a correlation of 0.15 for the quadriceps of young men (1985). Correlation was also weak following traumatic events or surgery (Leppilahti et al 2000, Mayer et al 1986, Renström et al 1983), in paraplegia (Calmels et al 1992) or

after phases of immobilization (Suzuki et al 1994). The limitations of the relationship between a structural measure of muscle mass and muscle performance are explainable not only by a slow structural adaption to short-term changes. Actual strength is dependent on structural and functional parameters, such as the actual muscle length, motor unit recruitment, shortening velocity, previous activity (Nigg and Herzog 2007) and sensory feedback (Hodges and Tucker 2011). Moreover, actual strength depends on psychological factors such as motivation and mood (Gendolla et al 2001). Even more complex, muscle activity may be static, dynamic, concentric or eccentric, isometric or isokinetic. Values to quantify strength depend strongly on the measuring conditions.

In spite of these limitations, CSA is an objective measure related to muscle mass and a recognized indicator for muscle wasting (Arokoski et al 2002), sarcopenia (Ochi et al 2010) and hypertrophy (Kanehisa et al 1998). Measurements of CSA are a static assessment that does not include the assessment of muscle activity itself.

2.5.4 Muscle thickness

The measurement of muscle thickness and its changes in activating muscle is the most common parameter to assess muscle function by US (Figure 2.5). Static thickness measurements represent a rapid, clinically feasible substitute for the more extensive measurements of anatomical CSA (Wait et al 1989), in particular on large muscles. CSA measurements require a sufficiently large field of view to image the complete cross-section of a muscle or the technical fusion of several images to allow for an extended view. The correlation of muscle thickness with CSA is good overall, although differences between muscles have been reported. Hides et al (1992) reported a correlation coefficient of 0.93 for the lumbar multifidus, Dupont et al (2001) reported a correlation of 0.96 for the deltoid and supraspinatus muscles, Rankin et al (2005) estimated the correlation between 0.66 and 0.89 for neck muscles, Hides et al (2006) found coefficients between 0.84 – 0.95 for transversus abdominis, O'Sullivan C. et al (2007) identified a correlation between 0.62 and 0.77 for the lower trapezius muscle and O'Sullivan K. et al (2009) observed a weak to moderate correlation of 0.18 – 0.59 for the quadriceps.

Muscle thickness, as muscle volume, as physiological and anatomical CSA, is a measure that serves as a structural correlate of a muscle's capacity to produce torque.

With each step of reduction for reasons of practicality, the correlation with muscle strength is less certain and more factors of influence are ignored (Figure 2.6).

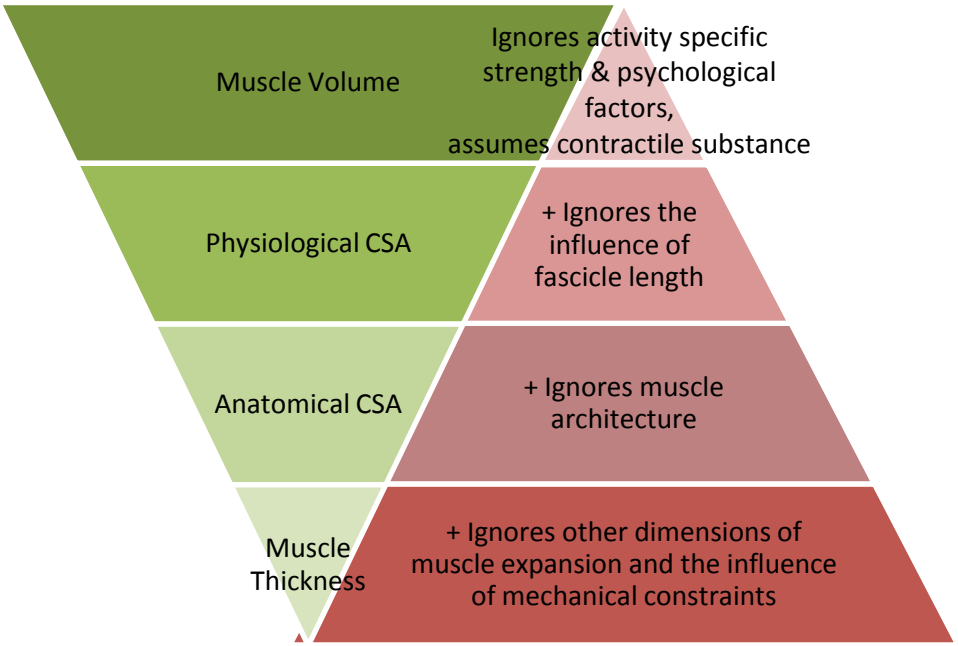


Figure 2.6. Structural correlates of a muscle's capacity to produce torque (green) and limiting factors (red).

In spite of these limitations, measurements of muscle thickness have been used to quantify muscle atrophy and training effects during ageing (Reimers et al 1998) and in neurologic diseases (Lee et al 2010, Ohata et al 2006). Muscle thickness measurements provided evidence for unilateral, local wasting of the lumbar multifidus in individuals with low back pain (Hides et al 1994) and were an outcome measure of transversus abdominis training in athletes with groin pain (Jansen et al 2009).

Muscle thickness is influenced by anthropometry, gender and handedness (Mannion et al 2008c). Thickness values need to be compared for interpretation. Methodological considerations, in particular when evaluating muscle thickness over time, include the reproducibility of the transducer position and angle (Whittaker et al 2009, Klimstra et al 2007) and comparability of different transducer types (Warner et al 2008). An important limitation of thickness measurements was demonstrated by Krentz and Farthing (2010) in their study of the effects of intensive, eccentric training on the biceps brachii. The authors measured an increase in thickness that had to be interpreted as an indication of swelling due to muscle damage and not as a sign of muscle activity or increased muscle mass.

2.5.4.1 Thickness changes in activated muscle

It is often assumed that muscles thicken during contraction and that the level of contraction can be assessed from the increase in thickness. Studies that compared changes of muscle thickness to the production of torque or the intensity of excitation in isometric contractions limit this assumption. Hodges et al (2003) demonstrated curvilinear relationships between EMG amplitude and muscle thickness increase for the biceps brachii and brachialis, the tibialis anterior, the obliquus internus and transversus abdominis muscles. No relationship was found for obliquus externus. Above 30% of maximal voluntary strength, there was little change in muscle thickness. In contrast, McMeeken et al (2004) reported a linear relationship between transversus abdominis excitation and thickness increase. The difference between Hodges' and McMeeken's findings may be explained by the different tasks. Hodges required maximal intraabdominal pressure for the 100% grade, whereas McMeeken used the maximal values of valsalva and abdominal hollowing manoeuvres. Possibly, the study of McMeeken et al included only the lower force range. Also limited to a range up to 50% of maximal effort, Kiesel et al (2007b) found a linear relationship between increase of lumbar multifidus thickness and EMG amplitude. Lee et al (2007b) observed a relatively linear increase in cervical multifidus thickness up to 50 % MVIC that flattened with higher force. John and Beith (2007) found no relationship and eventually even thinning of the obliquus externus muscle with increasing effort. Also Delaney et al (2010) reported reduced rectus femoris dimensions with increasing effort. In summary, evidence supports an increase in muscle thickness during isometric contraction in several but not all examined muscles. No prediction model proved consistently superior. Further limitations have been reported regarding fatiguing contractions. Several authors measured increasing muscle thickness during and following a fatiguing task (Shi et al 2007: biceps brachii, Ishikawa et al 2006: soleus, Bakke et al 1996: masseter).

The change of muscle thickness is related not only to the intensity of contraction but also to changes of muscle length. The influence of dynamic movement on muscle thickening has been evaluated only for some muscles. From these studies a different thickening behaviour for the concentric and eccentric phase has been reported (Xie et al 2009, extensor carpi radialis, Reeves and Narici 2003, tibialis anterior). Other researchers measured muscle thickness changes to predict a joint angle. Xie et al (2009) estimated accurately a one-dimensional wrist angle from

extensor carpi radialis thickening. Recent publications of this group report the development of a US controlled hand prosthesis based on the measurement of muscle thickening (Chen et al 2010, Guo et al 2009).This research group proposed the term 'sonomyography' analogous to electromyography (Shi et al 2008).

In spite of the stated limitations, important findings have been made using measurements of thickness change by US. For instance, Ferreira et al (2004) observed reduced transversus abdominis thickening in individuals with low back pain during an isometric leg task, Gill et al (2007) documented increased transversus abdominis thickening following spinal manipulation in a patient with low back pain, and Kiesel et al (2008) measured decreased thickening of transversus abdominis and lumbar multifidus after local pain provocation by injected saline. These reports refer to muscles for which the relationship between thickness and force had been established and negligible length changes were involved. Therefore, it could be concluded that acute and chronic pain result in decreased activity of selected deep trunk muscles and that transversus abdominis activity is improved following spinal manipulation.

On the contrary, studies of muscle thickness changes during tasks that include length changes of the target muscles are difficult to interpret (Teyhen et al 2008, discussion, Whittaker 2007). Therefore, comparing muscle thickening in different postures (Ainscough-Potts et al 2006) or during selected exercises (Endleman and Critchley 2008, Teyhen et al 2008) is explorative, but not clearly indicative of differences in the intensity of activity.

In summary, the change of muscle thickness is currently the most commonly used US measurement of muscle activity. Measurements of thickness change allow for distinguishing low levels of isometric activity in selected muscles for which the nature of the relationship between torque production and thickening has been proven.

The following section presents measurements of the architectural parameters of muscle, the arrangement of the fascicles. Measurements of architectural parameters during activation are an ongoing area of research aimed to understand and model the mechanics of different types of muscle activation and the interplay between muscle and tendon (Lieber and Fridén 2000, Narici 1999).

2.5.5 Architectural parameters of muscle

Differences in the composition of muscles have historically been observed in cadavers and have led to the anatomical differentiation of parallel and pennate muscles. Real-time US imaging has enabled the observation of muscle fascicles in-vivo and even during activation (Fukunaga et al 1997). US measurements of muscle architecture refer to the angle of pennation of the fascicles towards the tendon and to fascicle length. Both parameters have a functional significance for modelling a muscle's capacities to produce torque and to control range of movement (Lieber and Fridén 2000).

The terms 'fascicles', 'fibre bundles' and 'fibres' are often used synonymously (Chleboun et al 2007, Chow et al 2000), which is not strictly precise as it has been found that fibres taper within a fascicle so that fibre length does not equal fascicle length (Lieber 2010, pp. 32/33, Fukunaga et al 1997). US delineates fascicles (Walker et al 2004) or fascicle-like structures (Bénard et al 2009) but not single muscle fibres. The term 'fascicle' is derived from the Latin fascic = bundle, or fasciculus = small bundle. Literally, a fascicle is a fibre bundle. However, with regards to the functional interpretation of muscle architecture, the distinction between fibres and fascicles is not made (Lieber 2010, Herzog 2007b). Lieber (2010, p. 32) described that short fibres within longer fascicles are arranged in series to act effectively like a long fibre, an argument that the distinction of fascicles and fibres is not necessary for the functional interpretation. Here, the terminology of the cited literature is used without distinguishing fibre and fascicle.

There are two main physiological implications from muscle architecture. First, fibre length is proportional to the range the fibre can shorten (Lieber 2010, p. 34). The longer the fibres of a muscle, the more the muscle is designed to control a large range of motion. Second, the pennation of muscle fibres is a design for increasing the physiological CSA (PCSA) (Lieber 2010, p. 37) and therefore the capability to produce torque (Lieber 2010, p. 34). The interdependence of PCSA and pennation angle is indicated by the calculation of PCSA, in which MV is muscle volume, θ is the pennation angle and FL is fascicle length (Fukunaga et al 2001):

$$PCSA = \frac{MV \cos\theta}{FL}$$

Formula 2.3. Physiological cross-sectional area of a muscle (PCSA).

Ward et al (2009) measured the architectural parameters of 28 lower limb muscles. The muscle list did not include gluteus minimus. These data indicate a correlation of 0.72 between PCSA and pennation angle (Appendix 2). A higher pennation angle suggests a larger PCSA and a higher capacity to produce torque. If the average fascicle length and pennation angle of a muscle are known, conclusions on functional characteristics in comparison to other muscles can be drawn (Lieber 2010, p. 116-119, Narici 1999). In the following, gluteus medius architecture is ranked in comparison to 27 other lower limb muscles, from low to high values, based on the data of Ward et al (2009). An average fibre length of 7.3 cm and a pennation angle of 20.5° have been reported for gluteus medius. Of 28 lower limb muscles, tibialis posterior has the shortest and sartorius the longest fibres, 3.8 and 40.3 cm, respectively. With regard to fibre length and the ability to control range of motion, gluteus medius is ranked 12th comparable with extensor hallucis. Sartorius has the smallest and vastus medialis the largest pennation angle, 1.3° and 29.6°, respectively. With regard to pennation angle and the ability to create torque, gluteus medius is ranked 25th comparable to gluteus maximus. Similarly with regard to its PCSA, gluteus medius is ranked 26th comparable to gluteus maximus and vastus lateralis. Gluteus medius architectural parameters indicate the muscle's main function is torque production rather than covering range of motion (Ward et al 2010) (Appendix 2).

Muscle architecture has also been studied to investigate the structural influences of training (Potier et al 2009, Blazevich et al 2003), trauma (Bleakney and Mafulli 2002), disease (Matschke et al 2010, Li et al 2007a), disuse (De Boer et al 2008, Kawakami et al 2000) or ageing (Thom et al 2007, Narici et al 2003). Further, architectural studies build the base for complex muscle models (Blemker and Delp 2005).

2.5.5.1 The change of architectural parameters during activity

The static assessment as well as the dynamic behaviour of muscle fascicles in different types of contraction has been studied. During isometric muscle contraction, muscle fibres shorten and the pennation angle increases, both in a non-linear fashion (Hodges et al 2003, Maganaris et al 1998, Herbert and Gandevia 1995). During isokinetic, concentric activations, curvilinear changes of fascicle length and pennation angle have been observed, whereas in eccentric activation fascicles kept their length (Reeves and Narici 2003). The series elastic elements and tendon compliance mitigate

the contractile forces so that fascicle length does not replicate the motion occurring in the respective joint (Lichtwark et al 2007, Fukunaga et al 2002).

There are considerable methodological challenges for US studies on muscle architecture. The most important restriction is the visibility of the fascicles, which depends on the spatial resolution of the US system, the depth of the muscle, the thickness of the subcutaneous, adipose tissue and the possibility to take an approximately rectangular scanning approach towards the fascicle plane.
The fascicle plane needs to be known to measure the fascicles. Bénard et al (2009) found 14 - 23% error in measurements of fascicle length and pennation with a deviation of 15° of the scanning plane relative to the anatomical fascicle plane.
The fascicle is commonly not visible in full length. Its extensions are extrapolated from visible parts and its length is estimated, not measured (Bénard et al 2009, Li et al 2007a, Mairet et al 2006, Kubo et al 2003). This extrapolation is based on the assumption of straight fascicles, which is an idealization as fascicles are often curved (Bénard et al 2009, Wang et al 2009, Blazevich et al 2006).
Fascicle like structures which did not always represent true fascicles have been observed (Bénard et al 2009). This finding emphasizes the importance of precise knowledge of sectional anatomy and artefactual delineation.
Reproduction of the scanning plane for repeated measurements is a further issue in US imaging of muscle architecture (Whittaker et al 2009, Klimstra et al 2007), as it is difficult to control.
Fascicle parameters need to be measured not only at one but at several locations of a muscle to be representative. Significant differences between proximal and distal architectural parameters have been demonstrated (Kim et al 2010b: supraspinatus, Lichtwark et al 2007: gastrocnemius, Blazevich et al 2006: quadriceps femoris). These methodological difficulties complicate the use of architectural measurements.

Muscle thickness is strongly related to muscle architecture. When the pennation angle θ, the fascicle length and the angle γ between the lower and the upper fascia is known, muscle thickness can be calculated based on the assumptions of two-dimensionality and straight fascicles.

$$Muscle\ thickness = \frac{Fascicle\ length \times\ sin\ (180° - (\gamma + 180° - \theta))}{sin\ (\gamma + 90°)}$$

Formula 2.4. Calculation of muscle thickness from architectural parameters of muscle (Blazevich et al 2006).

This formula suggests that the change of muscle thickness during contraction depends on fascicle behaviour, explicitly on the amount of fascicle shortening in relation to pennation angle change.

The methodological challenges for US studies of dynamic muscle architecture prevent the broader use of architectural parameters in RUSI. However, knowledge of the dynamic fascicle behaviour may be important to interpret the change of muscle shape and muscle thickness.

Architectural parameters are measured in B-mode US. B-mode is the most commonly used US mode. B-mode provides a two-dimensional, greyscale US image of the sound reflecting interfaces in the scanned tissue section. The B-mode image is constructed from a tight sequence of vertical lines which are rapidly written one after the other. Each line corresponds to one interrogation of the tissue under the transducer. To produce the lines, the US beam must sweep rapidly along the footprint of the transducer. The sweep needs time, limiting the temporal resolution of B-mode and the temporal accuracy of the detection of motion.

To assess dynamic changes during activity, other US modes provide the option to observe motion in higher temporal resolution. Echocardiography is the US area with most experience in the observation of motion. The assessment of cardiac motion has been developed since the Fifties of the last century and initiated the sonographic discipline of echocardiography (Roelandt 2000). Attempts have been made to transfer typical echocardiography approaches to the assessment of motion during muscle activity (Grubb et al 1995). The following section introduces specific US modes for assessing tissue motion and the literature pertaining to this US application.

2.5.6 Muscle motion

It is difficult to track motion in B (Brightness)-mode US, because the tracking of reference points over time in consecutive frames is required (Shi et al 2008, Peng et al 2007). Other US modes that portray tissue displacement or motion velocity in the time domain are designed for observing motion, such as M (Motion)-mode or Doppler modes. In the following, the principles and specific advantages of M-mode and Doppler US modes for the assessment of motion during muscle activity are presented.

2.5.6.1 M-mode

B-mode and M-mode US images consist of vertical echo lines. In each line, sound pulses are sent into the body section under the transducer. When a sound pulse meets a reflective interface, sound is reflected back into the transducer. The received echoes are analysed for the time interval between sending and receiving, which is translated into the depth of the reflector. Further, the echoes are analysed for their intensity, which translates into the greyscale of the reflector's delineation in the US image.

The B-mode image consists of a dense sequence of single lines written in rapid sequence. The footprint of the US transducer contains a line of tightly packed sound-producing crystals that are activated in sequence to have a sweeping sound-beam to write these lines (Whittingham and Martin 2010). The M-mode trace also consists of a dense sequence of lines. However, the M-mode lines originate from a single crystal which is activated in rapid sequence to write a new line every few milliseconds about the current position of the interfaces in the sound beam (Anderson 2007, Gent 1997). M-mode US adds the dimension of time at the expense of the second dimension of the B-mode image. M-mode informs only about the depth and the echogenicity of an interface, not about its width. The advantage is a temporal display with high acuity to distinguish two consecutive events. The disadvantage is the loss of anatomic shape. Therefore, a B-mode image is added to the M-mode image for anatomic orientation (Figure 2.7).

The M-mode trace of muscle tissue displays horizontal lines of different greyscales. Hyperechoic lines originate from fibrous structures such as the periosteum or the perimysium. Hypoechoic lines arise from water-rich tissue (Walker et al 2004). A straight, horizontal line indicates a non-moving interface. When an interface moves the

line changes its greyscale or its position. The temporal resolution of M-mode US is high. For example, a temporal resolution of 3.8 ms in M-mode versus 31.25 ms in B-mode was achieved in the ANTARES US system, which was used in the surface EMG study of this research, and an M-mode resolution of 2.2 ms versus 15.4 ms in B-mode were given in the Xario US system, which was used for the fine-wire study.

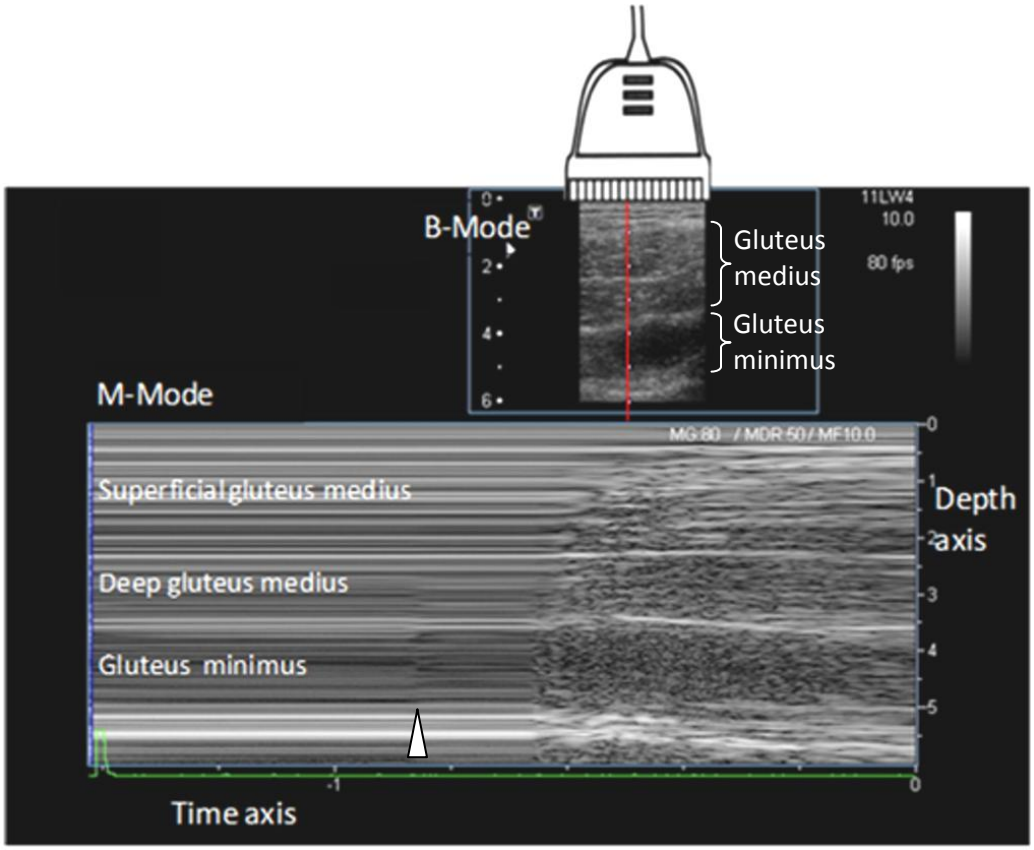


Figure 2.7. M-mode image of an isometric gluteus medius and minimus activity in 40% MVIC. The arrowhead indicates slight motion originating in the pulsation of a vessel.

2.5.6.2 Musculoskeletal applications of M-mode

One of the first reported observations of skeletal muscle motion by M-mode US refers to the diaphragm (Haber et al 1975), an application which is still in use (Kim et al 2010a). A further, early M-mode application relates to the documentation of fasciculation. According to Reimers et al (1988) fasciculation could be detected and distinguished from arterial pulsation (Figure 2.7), shivering, involuntary movements and incomplete relaxation by the M-mode pattern. This application is currently in use for the quantitative assessment of neurological disorders (Arts et al 2011). Physiotherapists

have applied M-mode for monitoring thickness change during activity of transversus abdominis (Bunce et al 2004, McMeeken et al 2004, Bunce et al 2002, Kidd et al 2002).

More recently, M-mode has been used to measure the onset of activation. Vasseljen et al (2006) compared the onset of tissue displacement in M-mode with the onset of electrical excitation for the lumbar multifidus muscle in 10 mm and 30 mm depth. The results pointed to a high agreement of tissue displacement and excitation onset with a small, systematic delay of 16 ms for tissue motion, but only for the first activating, deep muscle. In the later activating, superficial muscle, tissue motion started often before excitation and the discrepancy between both events was larger (Vasseljen et al 2006). A similar investigation in the abdominal muscles replicated these results (Vasseljen et al 2009). It was concluded that M-mode is able to inform about the onset of the first but not later activating muscle levels (Westad et al 2010). Motion transmission from the first activating muscle into adjacent tissue concealed excitation onset in later activating muscles (Vasseljen et al 2006).

The literature on M-mode applications on skeletal muscles distinguishes two principal types of M-mode analysis, one for the displacement of tissue boundaries (Anderson and McDicken 1999), typical for echocardiography, diaphragmatic motion and thickness measurements, and a second for the change in greyscale patterns, for example in the detection of fasciculation or onset of motion. The assessment of tissue boundaries is established and well recognized. The interpretation of greyscale patterns is a relatively unexplored area that needs further research.

2.5.6.3 Doppler modes

Doppler US modes are based on the frequency analysis of the echoes. A frequency shift between the emitted sound-pulse and its echo indicates motion of the reflecting interface and whether the motion is directed towards or away from the transducer. This phenomenon is the ‘Doppler effect’ (Hoskins 2010, pp. 84-86). Calculation of the motion’s velocity requires the angle between the transducer and the direction of motion to be known. With no Doppler angle set, the US system assumes an angle of 0° towards the line of movement, which is muscle thickening if the muscle lies parallel to the skin (Hoskins et al 2010, pp. 89-116, Gent 1997, pp. 243-250).

There are multiple Doppler modes, and it is beyond the scope of this review to provide an overview. Basic differences between the Doppler modes refer to whether spectral information is colour coded, as in Colour Doppler, or on a spectral display, as in Pulsed Wave (PW) Doppler; whether the sampling region is a preselected small gate (PW Doppler), a single echo line (M-mode Colour Doppler) or a two-dimensional region of interest (B-mode Colour Doppler); whether low or high frequencies are removed by filters (Colour Flow Imaging versus Tissue Doppler); and whether two sampling gates are compared (Strain Imaging). In Doppler modes, in addition to the artefacts encountered by any type of US imaging, Doppler-specific artefacts must be considered (Hoskins et al 2010, pp. 134-137, Desco et al 2002, Gent 1997, pp. 280-284).

In this research, the most basic Doppler mode, PW Doppler was used. PW Doppler offers a detailed delineation of spectral changes in a technically simple environment. Echoes from the 'gate', a preselected depth in a single line of echoes, are analysed for their spectrum. On the spectral display time is plotted on the *x*-axis and the velocity of motion on the *y*-axis (Figure 2.8).

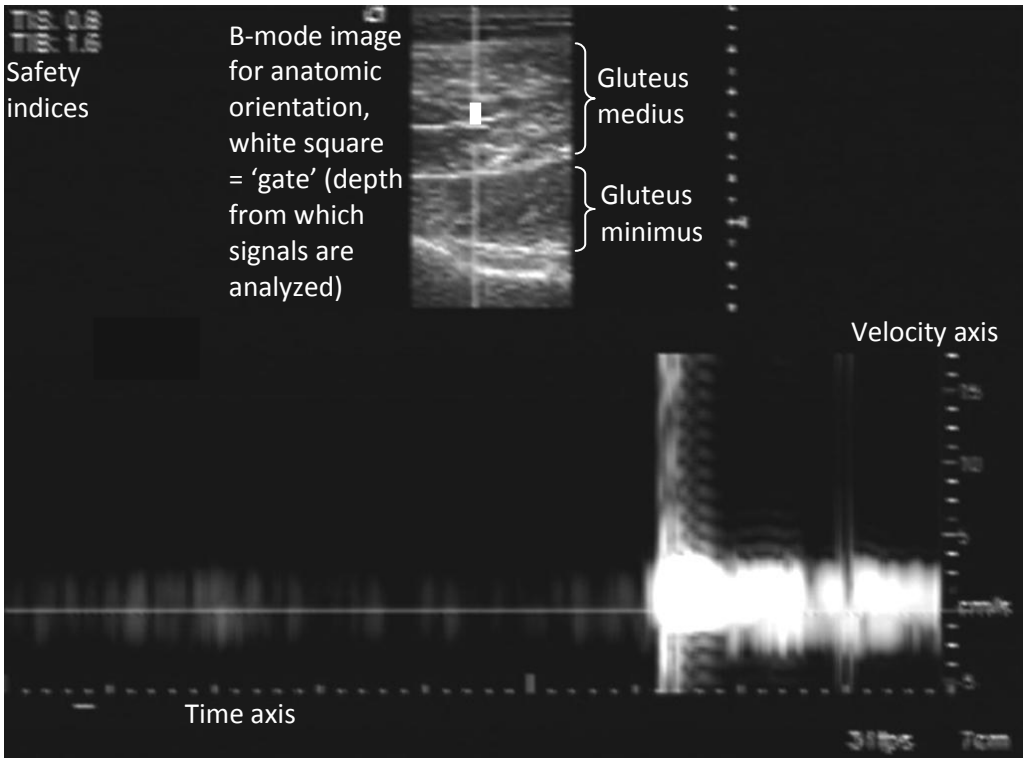


Figure 2.8. Pulsed-Wave Doppler, spectral display of an isometric gluteus medius activity in 40% MVIC.

Doppler imaging was originally established to gain information about blood flow. US signals of blood flow contain high-frequency velocities with low amplitude. In contrast, US signals of tissue motion contain low frequency velocities with high amplitude (Hoskins et al 2010, p. 95). For the purpose of assessing blood flow selectively, low frequencies are minimized by high-pass filters. In the mid-nineties, echocardiographers started to invert Doppler filters and isolate low frequency signals of tissue motion to describe myocardiac function (Fleming et al 1994). The new technique was termed Colour Doppler Myocardial Imaging, but later it advanced under the name of Doppler Tissue Imaging (DTI) (Palka et al 1995) or Tissue Doppler Imaging (TDI) (Mannion et al 2008a).

Muscles other than the heart were also assessed by TDI. Grubb et al (1995) reported the discrimination of passive and active rectus femoris and gastrocnemius motion, and the observation of different velocities during tendon reflexes using TDI. Further musculoskeletal Doppler applications included the measurement of tendon displacement (Holland et al 1999, Buyruk et al 1996) and of joint stiffness (Richardson et al 2002, Faber et al 2000, Buyruk et al 1995a, Buyruk et al 1995b).

More recently, TDI has been applied for detecting the onset of tissue motion during activity of the abdominal muscles and the quadriceps muscle (Mannion et al 2008a, Pulkovski et al 2008). Both studies reported a high correlation between tissue motion and excitation onset. However, onset could not be differentiated between different levels of depth, so that the TDI appeared to be not appropriate for comparing the onsets of adjacent muscles.

Using Doppler Strain Imaging, Vasseljen et al (2009) could identify the sequential onsets of excitation in the abdominal muscles by the thickening motion towards the transducer. Unfortunately, muscle thickening along the US beam was observed in only 50% of the subjects indicating a limited practical value of the application of Strain Imaging for detecting the onset of activity.

At the current stage, it can be stated that Doppler imaging detects tissue motion and its onset, but the relationship of motion and excitation is difficult to interpret and appears to be depth dependant. The exploration of a clinically meaningful assessment of muscle by Doppler US needs further research.

US imaging of tissue motion introduced long-time scanning of muscle activity with a fixed transducer (Bunce et al 2004). Long-time scanning leads to a relatively high, local US exposure. The risks of US imaging for this type of application should be reconsidered. The following section discusses the relevant side-effects of US imaging of muscle activity and states current safety guidelines.

2.5.7 Safety of the application of ultrasound imaging

The common sonographic procedure is to assess a relatively stationary area of interest by moving the transducer in different planes and angles. The physical impact of the US beam is focused on the object to be scanned. In the assessment of muscle activity, a stable transducer position must be secured while the scanned object moves. Assistive devices may be used to provide a secure, stable transducer position (Ishikawa and Komi 2008, Mannion et al 2008a, Lichtwark et al 2007, Bunce et al 2004, Bunce et al 2002). The physical impact of the US beam is strongest at the skin and subcutaneous tissue directly underlying the stationary transducer.

When the US beam travels through tissue, energy is propagated in the form of waves (Hoskins et al 2010, p. 181). US waves induce a temperature rise in the area of exposure, the thermal effect of US. Long-time imaging of muscle activity should be considered more hazardous than common diagnostic imaging because the same region is scanned over an extended time without moving the transducer. Limitations in the duration and power of an US application need to be considered.

If the scanned tissue contains gas, such as the lung or the intestines, the mechanical effect of US increases the risks of the US application. The oscillations of the US beam can induce the development and violent collapse of gas bubbles, the so called *cavitations*. This mechanical side effect of US has no relevance at the hip muscles and is not included in this review.

The appropriate use of medically applied US is regulated by national bodies which are strongly influenced by the standards prescribed by the International Electro-Technical Commission (IEC) (Duck 2007). The IEC did not set an upper limit for US output intensity. Instead, the information that US manufacturers are obliged to publish

on-screen has been specified. The responsible user must decide about the appropriateness of the application according to these thermal and mechanical indices, which allow for estimation of the risk of side-effects of the US exposure (Duck 2007).

In musculoskeletal imaging, the surface of bone is the tissue most likely to be heated. The TIB index (Thermal Index of Bone at focus) is most relevant (Hoskins et al 2010, 158-160, Duck 2007, p. 183). This index is an estimation of the temperature rise in the exposed tissue. A security factor is recommended due to large uncertainties about the real temperature rise in living tissue (Hoskins et al 2010, p. 184). The World Federation for US in Medicine and Biology (WFUMB) proposed a guideline according to which a temperature rise of up to 1.5° C is considered safe (Natori 2004). The following figures (2.9 and 2.10) indicate recommended scanning times with different TIB factors on the most sensitive foetal tissue and for general US applications on adults. The two figures indicate the range of recommendations depending on the sensitivity of the scanned tissue (Hoskins et al 2010, pp. 224/225) and may serve to designate the range within US exposure should be kept.

ALARA is the acronym for 'As Low As Reasonably Achievable' and refers to the main safety principle in the application of diagnostic US. US should be only applied when there is an expected benefit, the application should be completed in the shortest necessary time, US transmission must be interrupted ('freeze') when the image is not needed, the output power should allow the necessary penetration, but should not be higher (Gent 1997, p. 316).

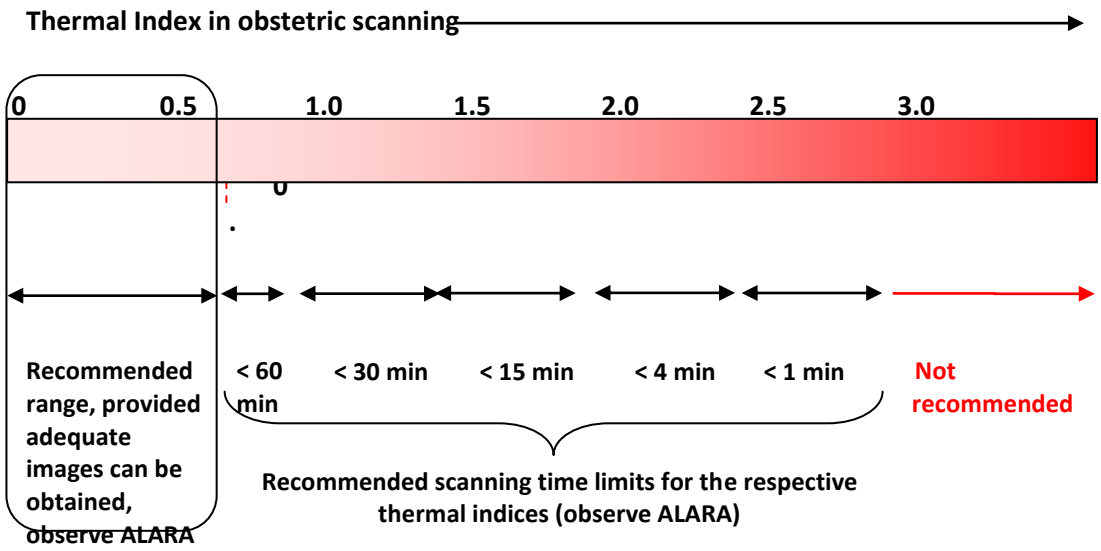


Figure 2.9. Recommended scanning durations in obstetric/prenatal scanning dependent on the thermal index (adapted fromBMUS 2009)

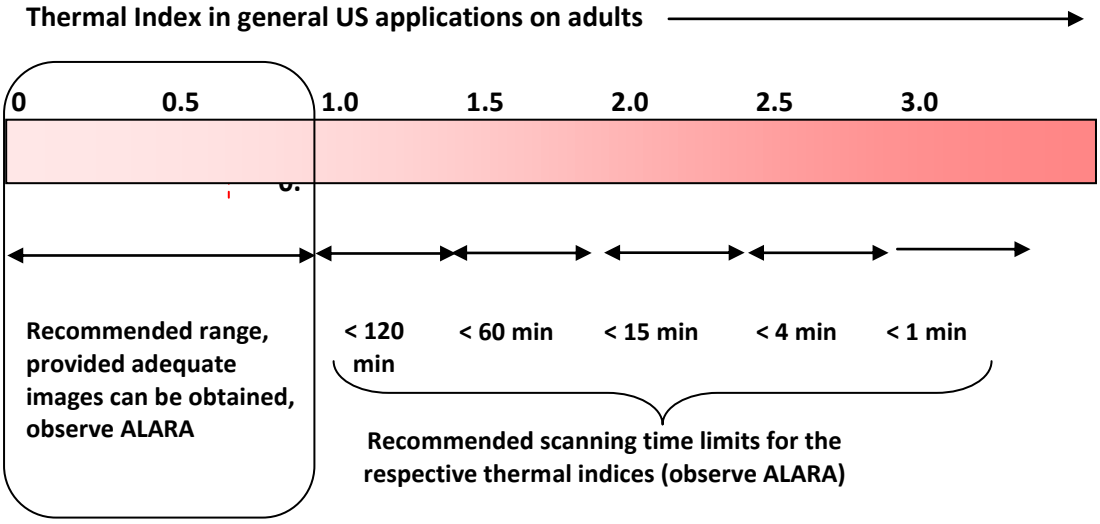


Figure 2.10. Recommended scanning durations in general abdominal and peripheral applications on adults dependent on the thermal index (Hoskins et al 2010, p. 225)

Up to this point, this literature review has provided the background for the assessment of muscle activity in this research. The following section presents the objects of investigation, the gluteus medius and minimus muscles at the lateral hip.

2.6 The Gluteus medius and minimus muscles

The gluteus medius and minimus muscles play a central role in controlling lateral stability not only for the hip joint (Grimaldi 2011, Grimaldi et al 2009b, Sims et al 2002) but also for the knee joint (Baldon et al 2009, Boling et al 2009) and the lower spine (Nelson-Wong et al 2008). Impaired function of the hip abductors contributes to lateral hip pain (Fredericson et al 2000), to patellofemoral pain (Prins and van der Wurff 2009) and to hip joint degeneration (Grimaldi et al 2009b, Arokoski et al 2002, Sims et al 2002). Gluteus medius and minimus function is commonly described by their abduction moment arm, modelled as a line of pull between their origin at the iliac bone and their insertion at the lateral and anterior facets of the greater trochanter (Gottschalk et al 1989). Beyond this obvious similarity of the gluteus medius and minimus muscles, several differences can be identified that point to unique aspects of gluteus medius or gluteus minimus function.

Figure 2.11. Anatomy dissection in which the middle part of gluteus medius has been removed to open the view on the gluteus minimus aponeurosis.

2.6.1.1 Moment arms

Studies in which moment arms were modelled for the anterior, middle and posterior gluteus medius and minimus separately reported the following differences. Gluteus medius is a stronger abductor than gluteus minimus (Neumann 2010, Pressel and Lengsfeld 1998). Gluteus medius is more effective as an external rotator than gluteus minimus (Neumann 2010, Delp et al 1999). The anterior part of gluteus minimus acts as a hip flexor, while the literature is inconsistent as to whether the anterior gluteus medius is effective to produce hip flexion (Neumann 2010, Dostal et al 1986). Gluteus minimus is more suitable to stabilize femoral internal and external rotation during motion in the sagittal plane as its rotational moment arms change less than that of gluteus medius (Delp et al 1999).

2.6.1.2 Muscle parts and detailed architecture

Reports of anatomy dissections describe the anterior, middle (or 'lateral superficial') (Jaegers et al 1992) and posterior parts of gluteus medius (Grimaldi 2011, MRI image, Gottschalk et al 1989). The functional division of gluteus medius is confirmed by the occurrence of differing onsets of activation of the sections found in EMG studies (Lyons et al 1983, Soderberg and Dostal 1978). In gluteus minimus, an anterior and a main, posterior part can be distinguished (Al-Hayani 2009). The anterior part formed a partly or completely separate muscle in seven of sixteen hips (Beck et al 2000), an observation that supports the separate function of the anterior gluteus minimus as a hip flexor.

Gluteus medius and minimus parts differ in fibre orientation (Figure 2.13) and pennation angle, and their fibres differ in length. In both muscles, the anterior fibres are orientated more vertically, whereas the posterior fibres run close to horizontal and almost parallel to the longitudinal axis of the femoral neck (Beck et al 2000, Jaegers et al 1992, Gottschalk et al 1989), which alludes to a joint-stabilizing function. Anterior gluteus minimus fibres may merge with gluteus medius (Al-Hayani 2009).

In gluteus medius, pennation angles of 20.4° ± 17.3° (Ward et al 2009) or 19°, 8° and 0° (Friederich and Brand 1990) have been measured. The difference between the studies may originate from measurements in different muscle parts or differences in methodology. Gluteus minimus pennation has been reported 10°, 1° and 0° (Friederich and Brand 1990).

Figure 2.12. Anatomy dissection of gluteus minimus.

Gluteus medius force is transmitted in a multipennate arrangement (Jensen and Davy 1975). Gluteus minimus fibres transmit their force by blending into the aponeurosis which thickens to form the gluteus minimus tendon (Beck et al 2000, Gottschalk et al 1989), a unipennate arrangement.

Ward et al (2009) found an average gluteus medius fibre length of 7.3 (± 1.6) cm, Friederich and Brand reported 5.4 to 8.4 cm (1990). Gluteus minimus fibre length was noted as approximately 4 cm by Beck et al (2000) and 3.8 to 6.8 cm by Friederich and Brand (1990). The greater length of gluteus medius fibres indicates a stronger emphasis on controlling hip abduction range of motion for gluteus medius compared to gluteus minimus.

Walters et al (2001) reported an additional insertion of gluteus minimus into the hip joint capsule. The authors concluded a protective function of gluteus minimus against entrapment of the hip joint capsule.

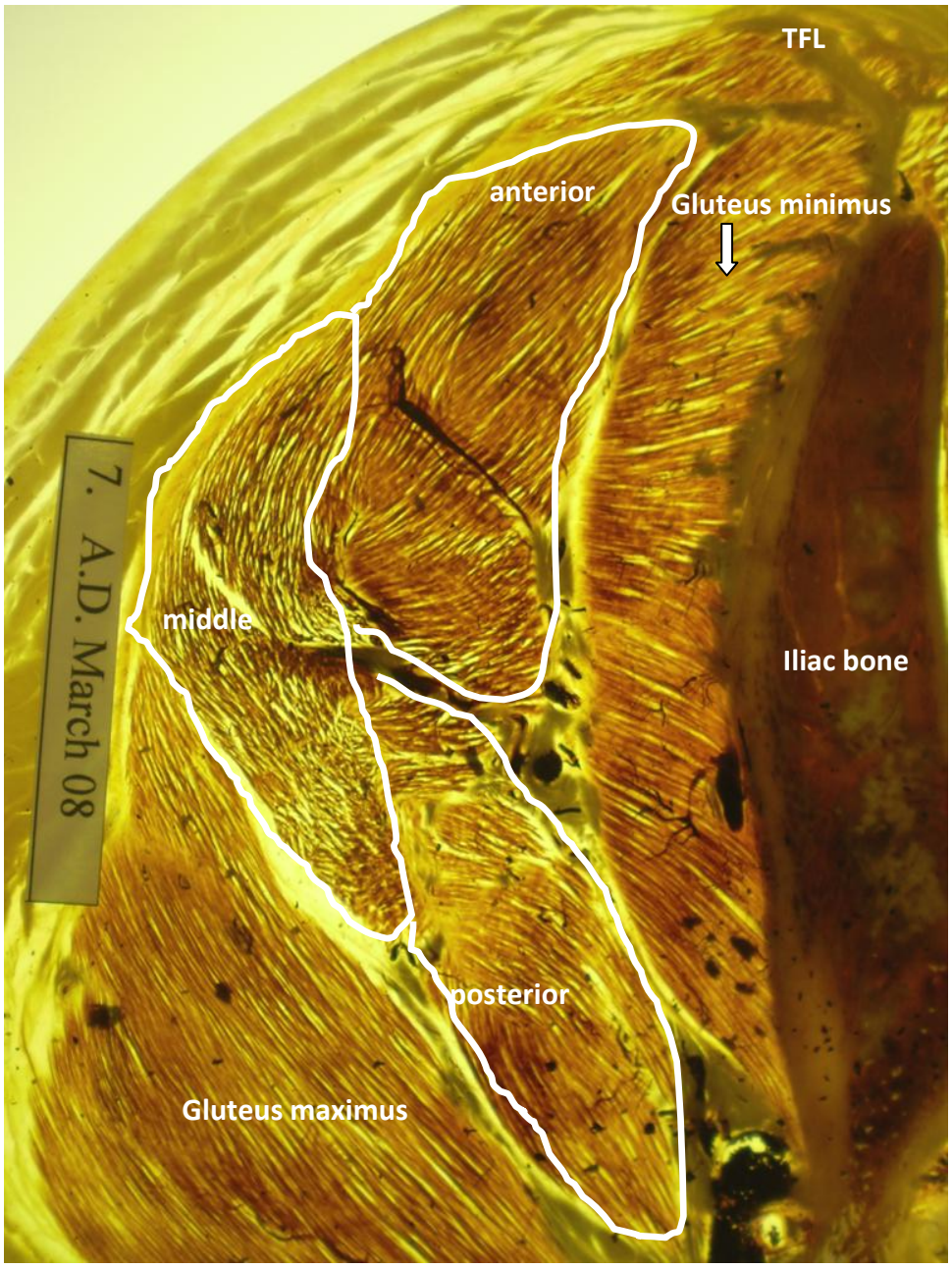


Figure 2.13. Plastinated section through the left lateral hip approx. 1 cm proximal of the greater trochanter. Gluteus medius parts (anterior, middle, posterior) differentiated according to differences in fibre directions.

2.6.1.3 Muscle composition

Gluteus minimus contains more than twice as many muscle spindles than gluteus medius (Stillman 2000), which designates a sensory motor function. The high content of slow twitch fibres in gluteus minimus, as shown in an animal study, gave rise to the assumption that its function is to sustain the neutral position of the hip joint, together with the piriformis, adductor brevis and quadratus femoris muscles (Hitomi et al 2005).

2.6.1.4 Suitability of gluteus medius and minimus for an assessment by electromyography and ultrasound imaging

The gluteus medius and minimus muscles at the lateral hip are readily accessible by EMG and US. The areas covered by gluteus maximus and the tensor fasciae latae can be identified from palpation and by US. The anterior muscle region offers an area free of large nerves and vessels for safe insertion of fine-wire electrodes (Schuenke et al 2007, p. 494, von Hochstetter triangle), and also sufficient surface to place an US transducer and surface EMG electrodes. Descriptions of US imaging standards to scan the gluteus medius and minimus muscles have been published (Ikezoe et al 2011, Bianchi and Martinoli 2007, pp. 571/572).

2.7 Summary and conclusions

Muscle activity includes electrical, chemical and mechanical processes. Muscle activity can be measured by the associated change in electric voltage, the displacement and motion of muscle tissue and the resulting torque. Each of these measurements reflects only one aspect of activity that is used to provide information on a complex procedure.

Measurements of activity by electrical excitation and by torque are well established, and their value has been extensively proven, in spite of inherent limitations. Measurements by US imaging appear suitable to overcome some limitations of dynamometry and EMG, in particular for measuring deep muscle activity selectively and non-invasively. However, in comparison to EMG and dynamometry, US measurements of muscle activity are little explored. Current literature does not report a simple US measurement that indicates the activity of a muscle unambiguously. In their current form, neither the thickening nor the motion of a muscle appears to be a universally applicable parameter for the assessment of muscle activity.

Further gaps identified in this review refer to the few muscles which have been assessed by US imaging and the limited knowledge on factors which influence and confound US assessments of activity. The review cannot clarify whether the patterns of hip abductor activity can be assessed by US imaging or which modes and measurements should be used. The following research contributes to filling the identified gaps.

3 Chapter Three: Assessing Hip Abductor Activity by Ultrasound Imaging: Development and Preparation of the Experimental Setups for Data Collection

Overview of Chapter Three

- Section 3.1 introduces basic, methodical decisions for measuring gluteus medius and minimus activity, e.g. the measurement location.
- Section 3.2 presents general considerations regarding the design and the objectives of the experimental setups.
- Sections 3.3 and 3.4 describe the detailed methodology of the two experimental setups which provided the data for the methodological part of this research, a setup including surface EMG and different US modes; and a second setup including fine-wire EMG and M-mode US.

3.1 Basic, methodological decisions before the application of EMG and ultrasound imaging for measuring hip abductor activity

The research aimed to compare the activity of the gluteus medius and minimus muscles in subjects with and without hip pain. Activity may be described by its level and its timing (Chapter Two, section 2.33). A pattern of activity may be described by the selective activity in relation to an external event (Steele and Brown 1999). The current standard method to assess selective muscle activity is EMG. Surface EMG is applicable on gluteus medius, but for gluteus minimus invasive fine-wire EMG is needed. Chapter Two, sections 2.3.4 and 2.3.5 stated the limitations of fine-wire EMG. Alternatively, gluteus minimus activity may be measurable using US imaging. However, no standards exist to assess gluteus medius or minimus activity by US. It was necessary to compare EMG and US measurements of gluteus medius and minimus activity to identify an applicable US assessment of hip abductor activity.

Some basic considerations and tests were required before a comparison of EMG and US measurements could be started. The first steps were the identification of a suitable US scanning technique to image the gluteus medius and minimus muscles and the verification of the muscles in the US image by comparison with an anatomy section.

3.1.1 Ultrasound scanning of the hip abductors: scanning location

Bianchi and Martinoli (2007, p. 570) proposed to scan the lateral hip in a side-lying position. Side-lying has the advantage that gravity reduces the adipose tissue at the lateral hip. Adipose tissue is a bad sound conductor, which deteriorates the resolution of US images. Side-lying is a difficult position for standardizing different levels of abductor torque. In a study which did not include measurements of muscle activity, Ikezoe et al (2011) proposed to scan the gluteus medius and minimus muscles in a prone position. For the study task of isometric hip abduction, the prone position appeared disadvantageous because leg rotation is not neutral and is difficult to standardize, the subjects cannot observe lower limb motion or the procedures happening around them. It was decided to examine hip abductor activity in the supine position against a dynamometer because femoral external / internal rotation could be

standardized, force levels lower than leg weight could be included and the level of exerted torque could be measured. To reduce adipose tissue being pressed into the scanning area, additional cushioning of the plinth with an excavation for the scanned buttock was provided.

The abductors insert into the anterior and lateral facets of the greater trochanter that serves as the bony orientation landmark to determine the scanning location. Both literature references (Ikezoe et al 2011, Bianchi and Martinoli 2007, p. 571) suggest longitudinal scanning of gluteus medius and minimus proximal to the greater trochanter on a line parallel to the midsagittal line. For US thickness measurements, the mid-belly of the muscle should be imaged as the largest change is expected to happen at the location of maximal muscle diameter. This US scanning location is the same as the recommended position for EMG electrodes (Rainoldi et al 2004, Freriks et al 1999).

On gluteus medius, alternative EMG electrode positions could be ventral or dorsal to the US transducer. The position of the greater trochanter relative to the iliac bone is variable, and depends on the antetorsion angle of the proximal femur. Antetorsion angles differ between individuals. In a retrotorted femur, the greater trochanter is lateral and dorsal, and scanning gluteus medius on a line parallel to the midsagittal line would only leave space for surface electrodes ventral to the transducer. On the contrary, in a subject with an antetorted proximal femur, gluteus medius scanning parallel to the midsagittal line would be close to tensor fasciae latae, only leaving space for surface electrodes dorsal to the transducer. For a standardized location for the surface electrodes, the longitudinal scanning position needed to be oblique (Figure 3.1).

An oblique scanning location leaves space for the surface electrodes dorsal to the US transducer on the recommended height of the gluteus medius muscle or cranial to the US transducer on the same muscle part as the transducer (Figure 3.2). The cranial electrode position near the gluteus medius origin possibly results in a weaker EMG signal (De Luca 1997). However, for the comparison of activity onsets, the same muscle part should be measured because onset differences between gluteus medius parts have been reported (Soderberg and Dostal 1978).

The US scanning location was determined on the lower half of a reference line that connected the tip of the greater trochanter to the anterior quarter of a line between the anterior superior iliac spine (ASIS) and the posterior superior iliac spine (PSIS) (Figure 3.1). Surface electrodes were attached dorsal and cranial to the foam block in which the US transducer was fixed to the pelvis (Figure 3.2). This scanning location allowed for standardized measurements irrespective of individually different trochanter positions, and for measuring surface EMG close to the recommended electrode position and alternatively on the muscle part which was scanned by US.

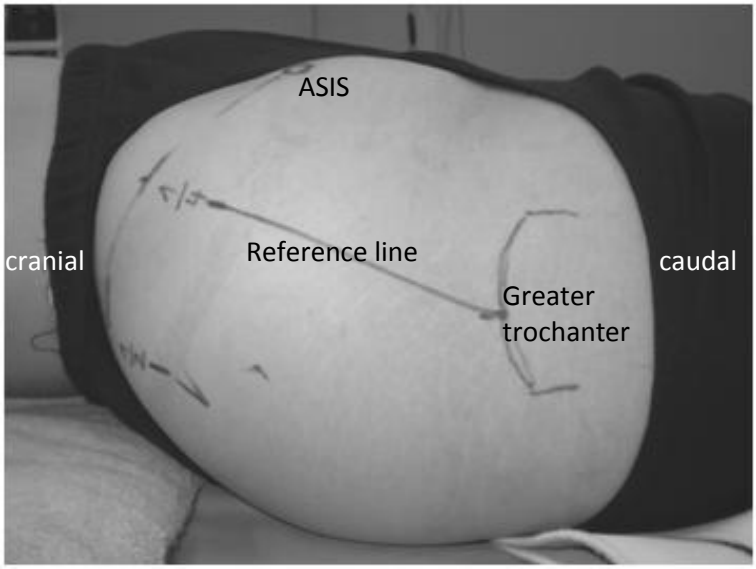


Figure 3.1. Subject is supine wearing a research pant which is cut out at the lateral hip to provide access to the gluteus medius and minimus muscles. The US transducer was positioned on the lower half of the reference line.

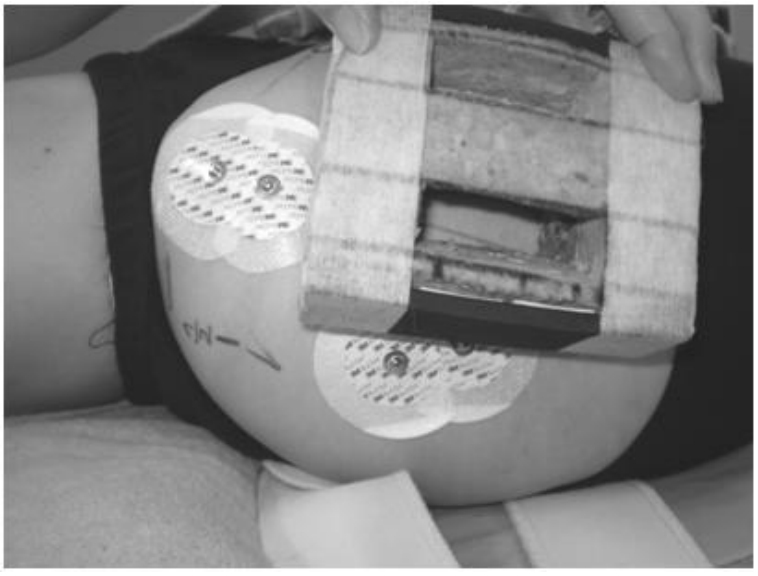


Figure 3.2. Surface electrodes dorsal and cranial to the foam block in which the US transducer was housed for fixation.

The next step was to identify the structures visible in the US image at the scanning location.

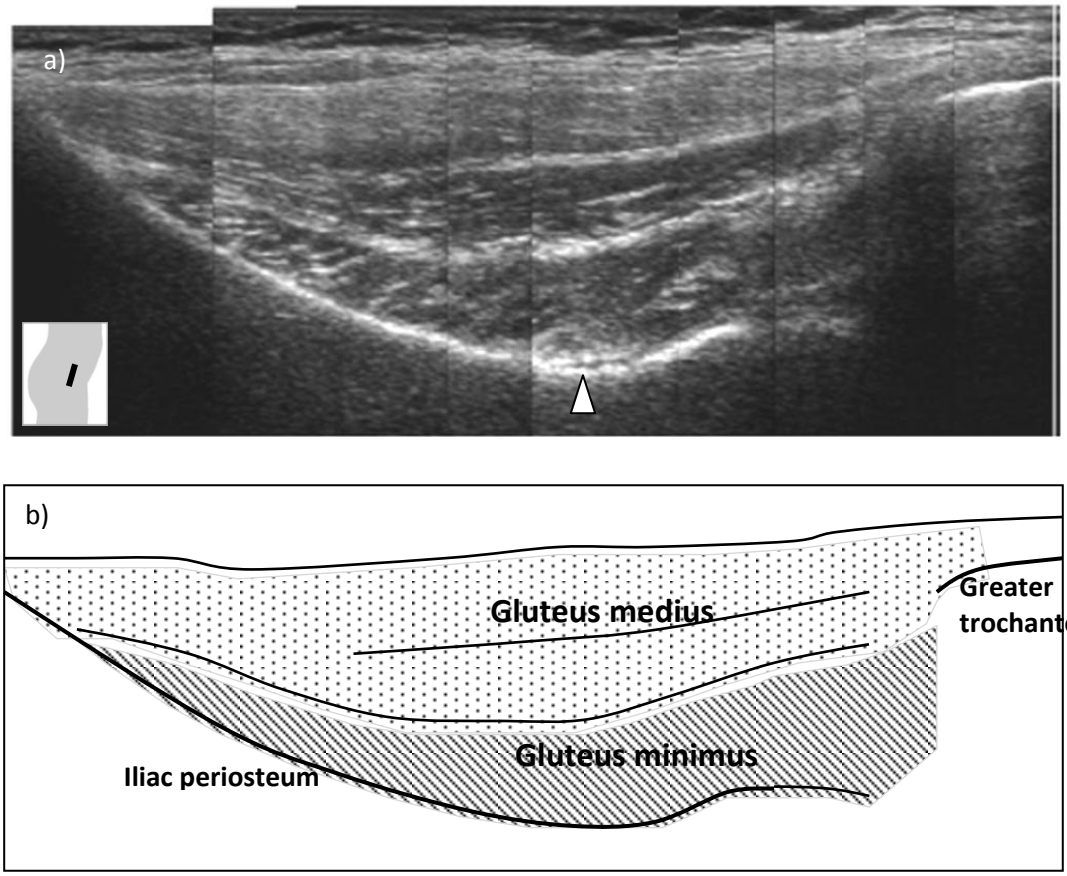
3.1.2 Identification of the gluteus medius and minimus muscles on the US image

A visual comparison of a collated US view with an anatomy section facilitated the identification of the gluteus medius and minimus muscles (Figures 3.3 and 3.4).

The iliac periosteum forms the deep border of gluteus minimus and is the deepest interface in the image (Figures 3.3 a). The gluteus minimus aponeurosis forms a hyperechoic band that sets a continuous demarcation towards gluteus medius. Above the aponeurosis lies the thick bulk of gluteus medius, which may be divided by a hyperechoic, thin, not necessarily continuous structure. In dynamic imaging, muscle motion suggests that this structure is a tendon within the muscle. Also in dynamic imaging, the superficial fascia of gluteus medius is easily identified, whereas in static images the upper fascia tends to blend into the fascia lata. In the caudal direction gluteus medius and gluteus minimus taper into their tendons that can be followed towards the greater trochanter.

Following the periosteum towards the hip joint, a hyperechoic structure between gluteus minimus and the iliac periosteum can be identified (Figures 3.3 a). This structure is not clearly identifiable on a midsagittal section (Figure 3.4), but becomes more evident towards an anterior view. It is suggested that this structure represents the reflected head of the rectus femoris muscle (Naňka et al 2003).

To differentiate the tensor fasciae latae muscle, the transducer has to be turned 90° to transverse scanning. In the anterior part of the iliac bone tensor fasciae latae is identified in a triangular shape, superficial to gluteus medius with a clear fascial border to the gluteal abductors (Figure 3.5).



Figures 3.3 a) and b). Collated longitudinal US view on the gluteus medius and minimus muscles at the proposed scanning location. Arrowhead points at a structure which presumably is the reflected head of rectus femoris.

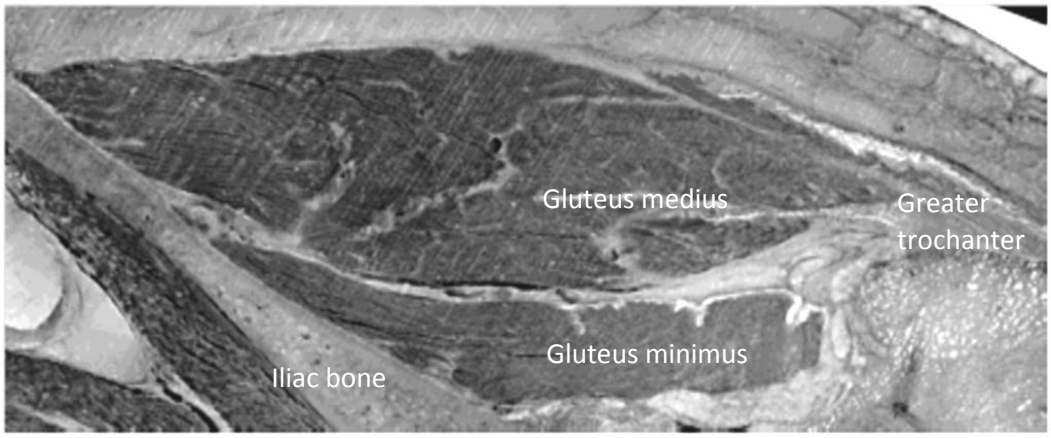


Figure 3.4. Anatomy section of gluteus medius and minimus, frontal plane. Reprint with kind permission of Primal Pictures Ltd., www.primalpictures.com, 1 October 2010 (Appendix 3.1).

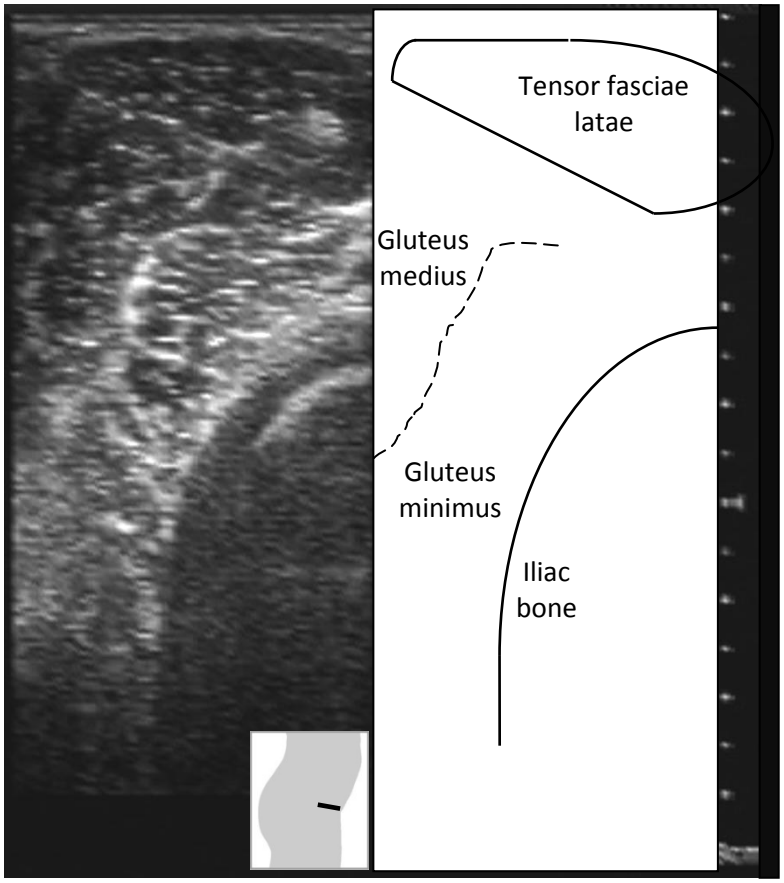


Figure 3.5. Transverse image 4 cm caudal to the ASIS. The tensor fasciae latae muscle is a clearly demarcated, oval muscle anterior and on top of the gluteal abductors. Anterior, gluteus medius and minimus may merge without clear demarcation (Bianchi & Martinoli, 2007).

Having identified the attachments of the EMG and US probes to examine hip abductor activity, the next question was whether the operation of the US transducer in direct vicinity to the surface electrodes creates interference in the EMG output.

3.1.3 Does ultrasound imaging create electromagnetic interference in surface EMG signals?

Electromagnetic interference is the degradation of electromagnetic signals, in this case by the potentially influencing electrical circuits of the US system. The EMG manufacturer assured a low effect of adjacent US on the EMG signals (Appendix 3.1).

The EMG output was examined in the experimental setup for interferential signal content by a visual comparison of raw EMG signals with and without an adjacent operating US transducer and by a frequency analysis of the signal content.

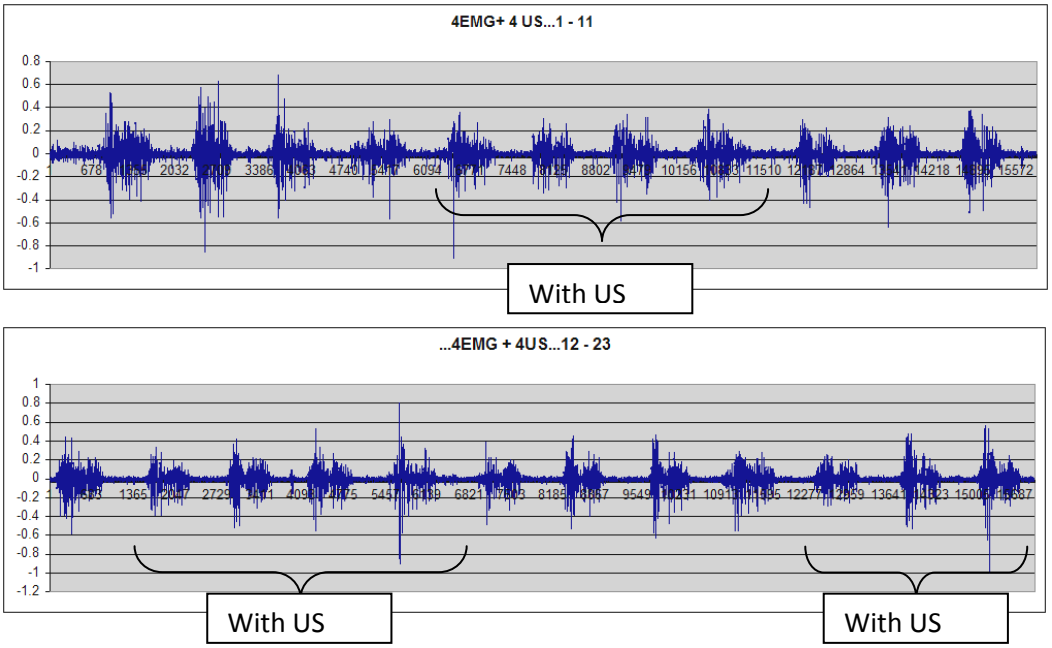


Figure 3.6. Gluteus medius activity while walking on the spot. Four trials with and four trials without an operating US transducer were recorded beside the EMG electrodes.

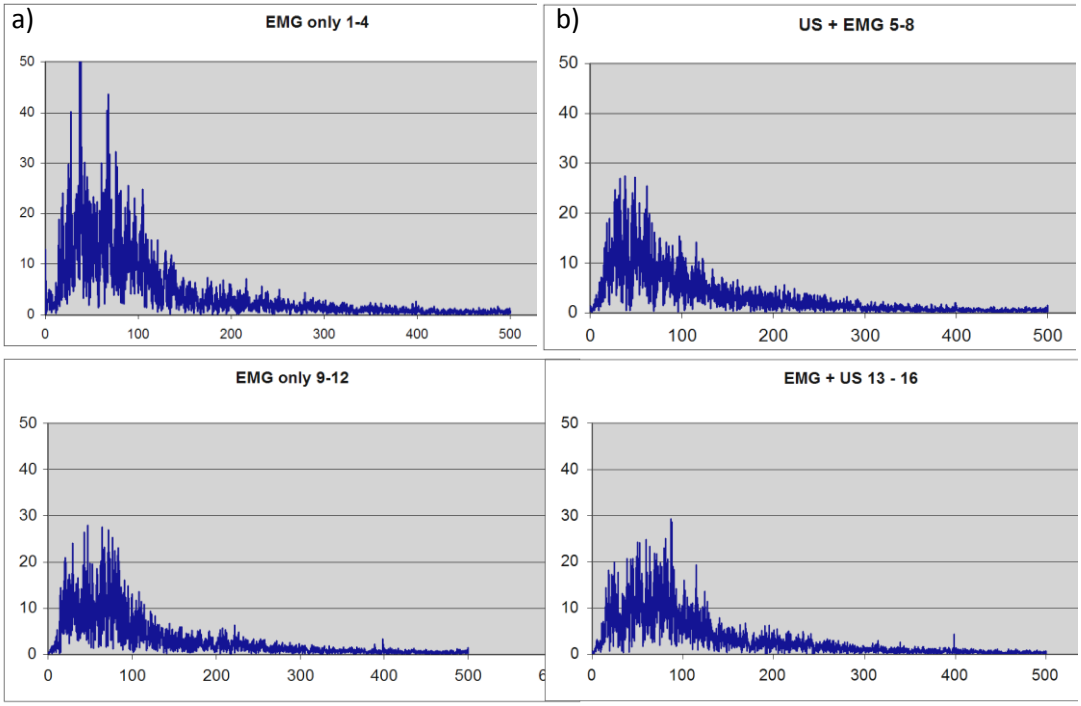


Figure 3.7. Frequency analysis (Fast Fourier Transform) of the initial four seconds of the first two series of walking on the spot without (a, c) and with (b, d) operating US transducer beside the EMG electrodes.

In the test setup, a subject stepped on the spot while gluteus medius activity was recorded by surface EMG. After four steps the powered US transducer with contact gel was positioned beside the EMG electrodes for the next four steps. Alternating, four steps with and without US transducer were recorded. Of the recorded 32 steps, 23 are documented in Figure 3.6. Raw signals were inspected for differences, which were not observable.

EMG signals of the initial four seconds with and without US were analysed for their spectral content (Figure 3.7). Electromagnetic noise was expected to produce irregular spikes within the frequency content of the signal (Chapter Two, Figure 2.1) and a shift in the median frequency. Additional noise spikes were not detectable. Analysis of the median frequency indicated no frequency shift in the four samples.

Having assured the feasibility of adjacent EMG and US data recording, the dynamometer was calibrated (Chapter Two, section 2.2).

3.1.4 Calibration of the dynamometer

Calibration means to compare measured values with known, valid reference values. Calibration verifies measurement validity irrespective of unknown influences that might have impact on the measurement instrument in time periods when the instrument is not controlled (Deutsches Institut für Normung 1995b, DIN 1319).

The dynamometer was calibrated with a set of normed weights. Linearity and repeatability of measurements irrespective of preloading were assured. The crossing point with the y-axis close to zero was documented. The documentation was needed to program the LabView application for force feedback.

With the procedures described in sections 3.1.1 to 3.1.4 several conditions for starting the data collection were established.

3.2 The design of the experimental setups

The overall objectives of this research were

- to identify or develop US measurements of the level and onset of gluteus medius and minimus activity, and
- to compare the patterns of gluteus medius and minimus activity between subjects with hip pain and controls.

Three setups for data collection were planned. The term *setup* instead of *study* is used to distinguish the events of data collection from the studies in which the data were analysed to answer study questions. Setups One and Two provided the data for Studies One to Five.

- Setup One included dynamometry, surface EMG (only gluteus medius) and US B-mode, M-mode and Pulsed-wave Doppler measurements in longitudinal and transverse scanning. The setup was a non-invasive data collection on asymptomatic volunteers designed to enable the estimation of the reliability of EMG and US measurements; to determine the relationship of EMG and US measurements of gluteus medius activity using different US modes and planes; and to determine the relationship of dynamometry and US thickness measurements of gluteus minimus activity.

- Setup Two included dynamometry, fine-wire EMG and M-mode US measurements in longitudinal scanning. The setup was an invasive data collection in order to deepen and extend the method comparison between EMG and US. The setup was designed to enable the comparison of EMG and US thickness and onset measurements of deep gluteus medius and minimus activity in supine and weight-bearing.

- Setup Three included the application of the relevant US measurements in a non-invasive study on subjects with hip pain and controls. M-mode US imaging was applied to compare the activity patterns of gluteus medius and minimus during step down.

In the following sections of this chapter, the detailed methods of data collection in Setups One and Two are documented. Data processing and analysis are described later with the respective studies. The methodology of Setup Three, the case-control study, is reported in Chapter Eight.

3.3 Setup One: isometric hip abduction recorded by surface EMG, B-mode, M-mode and Pulsed Wave Doppler US, longitudinal and transverse scanning planes

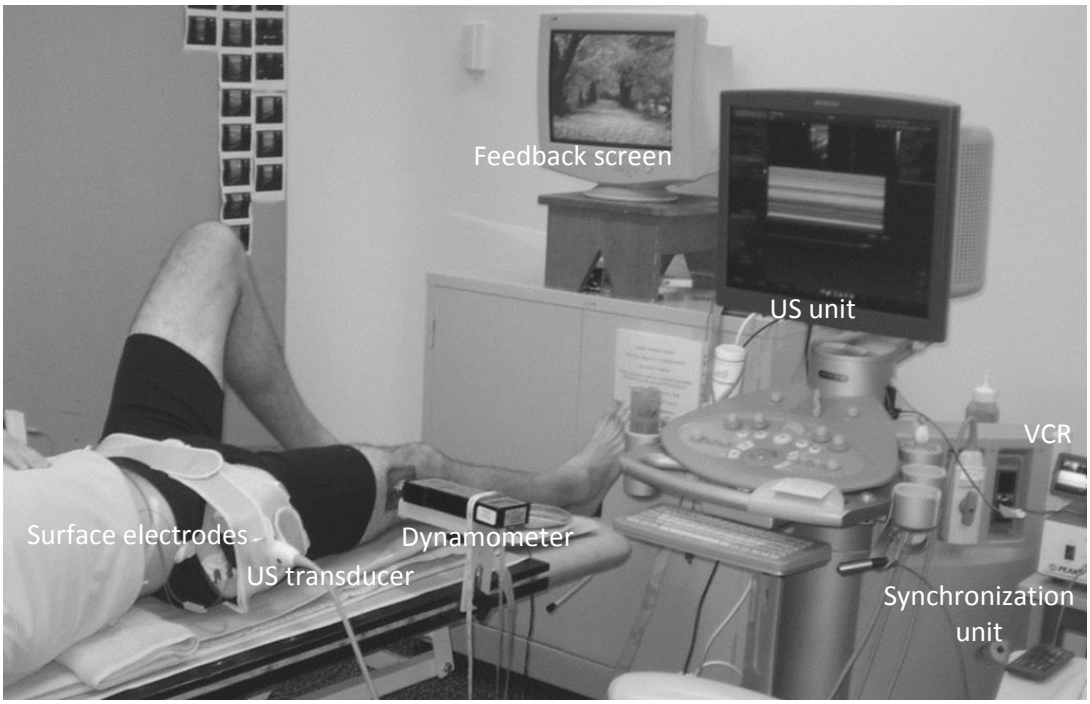


Figure 3.8. Setup One: isometric hip abduction recorded by dynamometry, surface EMG and US.

3.3.1 Subjects

A sample of convenience was recruited at the School of Physiotherapy, Curtin University. Exclusion criteria were hip pain, history of hip pathology or general musculoskeletal disease and BMI > 32 kg/m^2, as adipose tissue complicates US imaging. Sample size calculations based on published results of muscle latency studies (Stensdotter et al 2006, Hess et al 2005, Hodges et al 1996) indicated the need for sixteen subjects to detect a difference of 50 ms with a standard deviation of 50 ms per group. Ethics approval was obtained from the Human Research Ethics Committee,

Curtin University (PT0107/2007). Signed, informed consent was obtained in the two weeks prior to data collection. Appendix 3.2 documents the subject information and Appendix 3.3 the consent form.

3.3.2 Task and procedure

The task was isometric right hip abduction against a dynamometer. The simple task allowed for high standardization of task performance and graded activity in precise force levels. Activation is influenced by prior activation, an effect termed history dependence (Herzog 2007a, pp. 209-216). To minimize the influence of history dependence, a series of abduction trials in randomized force levels with fixed relaxation breaks was programmed in a LabView application (V8.2.1, National Instruments, Texas). Isometric abduction in four force levels of 20%, 40%, 60% and 80% of maximal voluntary isometric contraction (MVIC) had to be sustained for four seconds and was followed by relaxation breaks of 10 sec, 20 sec, 30 sec and 45 sec, respectively. These relaxation times had been tested to be sufficiently long to relax fully and sufficiently short to avoid boredom and distraction. Data recording included five series with different US modes and scanning planes (Table 3.1).

Table 3.1. Testing sequence in Setup One.

	Force	EMG	US	US plane	Trials
1.	Force	EMG	B-mode and M-mode	longitudinal	MVIC test, 2 repetitions recorded in B-mode + 1 in M-mode
2.	Force	EMG	M-mode	longitudinal	3 trials 20%, 3 trials 40%, 3 trials 60%, 2trials 80% in random order
3.	Force	EMG	PW Doppler	longitudinal	3 trials 20%, 3 trials 40%, 3 trials 60%, 2trials 80% in random order
4.	Force	EMG	M-mode	transverse	3 trials 20%, 3 trials 40%, 3 trials 60%, 2trials 80% in random order
5.	Force	EMG	PW Doppler	transverse	3 trials 20%, 3 trials 40%, 3 trials 60%, 2trials 80% in random order

Isometric right hip abduction in supine and from neutral joint position was performed against a dynamometer that was fixed to the side of the plinth, adjustable in position and height. Subjects were positioned parallel to the right side of the plinth with the hip joint in neutral position and with contact of the distal, lateral thigh to the dynamometer. A plate for stabilization was fixed on the left side of the pelvis. Subjects

were instructed to abduct the leg without flexing the hip, to keep the heel in contact with the plinth and the foot pointing vertically to the ceiling. Practice trials were repeated until the performance was correct. Then, the procedure followed the pre-programmed recording sequence. Subjects were informed about the exerted force and the required force level by a LabView programmed force feedback. Appendix 3.4 documents the study protocol.

3.3.3 Instrumentation

3.3.3.1 Dynamometry

Isometric abduction force was measured as the rise of pressure of the right thigh against the dynamometer (Mecmesin AFG-500N, Slinfold, UK). Signals were collected at 1000 Hz and low-pass filtered with a zero lag 4^{th} order Butterworth filter at 40 Hz.

3.3.3.2 Electromyography

EMG was recorded on an Octopus AMT-8 EMG system (Bortec Electronics Inc., Calgary, Canada), input impedance 10 GOhm, common mode rejection of 115 dB at 50 Hz with a sampling rate of 1000 Hz, bandwidth of 10 – 1000 Hz, using pre-gelled, round Ag-AgCl surface electrodes (3M Australia, Pymble, NSW) with a diameter of 8 mm recording surface and an inter-electrode distance of 22 mm. The ground electrode was attached to the lateral ribs. The skin was prepared for electrode attachment by shaving if necessary, slight abrasion and alcohol cleansing. After drying of the alcohol, the electrodes were attached on two locations (section 3.1.1 and Figure 3.2). One electrode location was in muscle fibre direction around the midpoint of a line connecting the greater trochanter to the iliac crest (Freriks et al 1999) dorsal to the foam block in which the US transducer was housed. The second electrode location was on the reference line for the US transducer cranial to the foam block. The first location was on the recommended muscle height but on a muscle part posterior to the muscle part scanned by US. The second electrode location was on the same muscle part as the US probe but more cranial than recommended (Freriks et al 1999). Raw EMG signals were pre-amplified 500 times, amplified, fed into a computer with an input amplitude range of + 10 V, 16 bit digitized and displayed on-screen (LabVIEW V8.2.1, National Instruments, Texas).

3.3.3.3 Ultrasound imaging

The abductors were imaged with an Antares 4.0 system (Siemens Medical Solutions, USA), a 49 Hz linear probe (VFX9-4) set to 9 Mhz, 38 mm footprint. The US system was set with the power necessary to penetrate the abductor muscles to the depth of the iliac periosteum. The right side of the image was orientated towards the greater trochanter. A single focus zone was set to the gluteus minimus aponeurosis.

In B-mode, a custom application was programmed to delineate high contrast for distinguishing the hyperechoic collagenous interfaces against hypoechoic contractile tissue, with low persistence to avoid blurred images by motion.
The M-mode settings included a small B-mode image for anatomic orientation on top of the M-mode trace. The M-mode trace represented 1.4 seconds of tissue motion at highest sweep speed. The sampling rate was, depending on the image depth, approximately 263 Hz, providing a temporal resolution of 3.8 ms. The M-mode beam was set slightly cranial to the reflected head of rectus femoris transecting the thickest muscle bulk (Figure 3.3 a).
Doppler US data were recorded with a pulse repetition frequency of 1221 Hz, a gate width of 5 mm, and the lowest wall filter setting (37 Hz). It was ensured that the Doppler gate stayed within gluteus medius during activity. Gain was regulated to display no or minimal signal in relaxation.

To reduce artefacts by transducer motion and secure the scanning angle over the full recording sequence, the transducer was housed in a foam block excavated at a 20° angle and fixed with belts around the pelvis (Mannion et al 2008a, Bunce et al 2002). The transducer angle, gain, time-gain-control (TGC) and dynamic range were adjusted to achieve a B-mode image that allowed for clear recognition of the iliac periosteum and the gluteus minimus aponeurosis and, if achievable, the recognition of single muscle fascicles.

US recordings of the complete testing procedure were captured at 25 frames per second onto video (Panasonic digital video camera, NV-MX 500A, Secaucus, USA) and exported to a computer for further analysis (iMovie, version 5.0.2).

3.3.3.4 Data synchronization

Synchronization between dynamometry, EMG and US data was accomplished by a manual trigger which started the dynamometer and the EMG and created a time

stamp on the concurrent video frame (Event Synchronization Unit, PEAK Performance Technologies Inc., Centennial, CO, USA).

3.3.4 Questions to be answered from the data collected in Setup One

Are US measurements of gluteus medius and minimus thickness and onset reliable?
The objectives to estimate the within and between days intra-tester reliability of gluteus medius and minimus *thickness* measurements by *B-mode* and *M-mode* US, and to compare the intra-tester reliability of gluteus medius *onset* measurements by EMG, *M-mode* and *Pulsed-Wave Doppler* US are examined in Chapters Four and Six.

US measurements of gluteus medius activity: Is the agreement with EMG higher in the longitudinal or the transverse scanning plane?
The objectives to compare the agreement of EMG *and thickness* measurements by *M-mode* US in the longitudinal and the transverse scanning planes, and to examine the relationship of EMG*, M-mode and Pulsed-Wave Doppler onset* measurements are investigated in Chapters Four and Six.

Is the level of abductor activity predictable from US measurements of gluteus medius and minimus thickness change?
The objective to quantify the prediction of the force level of isometric hip abduction, as measured by dynamometry and EMG amplitude, from the change of gluteus medius and minimus *thickness* was examined in Chapter Five.

Is the onset of electrical excitation of gluteus medius predictable from US measurements of motion onset?
The work on the objective to determine the agreement of measurements of gluteus medius activity *onset* by surface EMG, *M-mode* and *Pulsed-Wave Doppler* US is presented in Chapter Six.

Do EMG and US indicate differential activity of gluteus medius and minimus?
The objective to compare the level and timing of activity of gluteus medius to gluteus minimus was investigated in Chapter Six.

3.4 Setup Two: hip abductor activity in isometric abduction and step down recorded by fine-wire EMG and M-mode US

3.4.1 Modifications of the methods of data collection for the application of fine-wire EMG

Ethics approval for the fine-wire EMG study required the minimization of possible risks associated with fine-wire EMG, e.g. pain, vegetative reactions, bruising, infection and fine-wire breakage (Appendix 3.5). To reduce the risks, the method of data collection used in Setup One had to be modified.

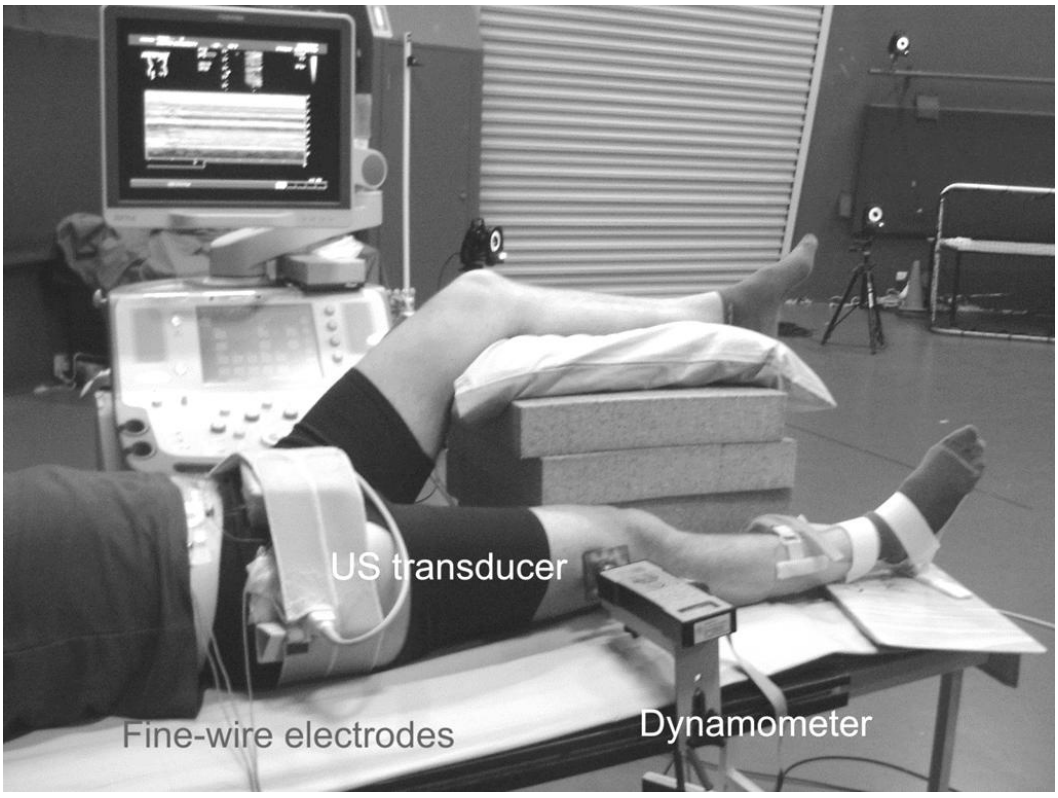


Figure 3.9. Setup Two: isometric hip abduction recorded by dynamometry, fine-wire EMG and US.

3.4.2 Subjects

Sample size calculations indicated the need for ten subjects to detect a difference of 150 ms with a standard deviation of 115 ms (based on the results of the Study Four, Chapter Six). To evade possible conflicts for dependent subjects, e.g. students in an invasive study, recruitment was among staff older than 30 years

(Appendix 3.6, subject information, Appendix 3.7 consent). Exclusion criteria were skin irritability in the hip or trunk region, previous hip surgery, pathology of the nervous system, medications potentially affecting the nervous system or reaction time, BMI > 32 kg/m^2. Ethics approval was provided by the Human Research Ethics Committee of Curtin University (HR 143/2009). Subjects signed informed consent.

3.4.3 Scanning location

The measurement location was the 'von Hochstetter triangle' (Schuenke et al 2007, pp. 494/495, Mensdorf 1999) (Figure 3.10), an area caudal and lateral to the anterior superior iliac spine and void of larger nerves and vessels. This area was selected to avoid bruising and irritation of nerves by insertion of the fine-wire electrodes (Mensdorf 1999, Schmidt 1957). The von Hochstetter triangle is situated slightly cranial and anterior of the measurement location in the surface EMG study. Figure 5.1 (Chapter Five) indicates both measurement sites for comparison.

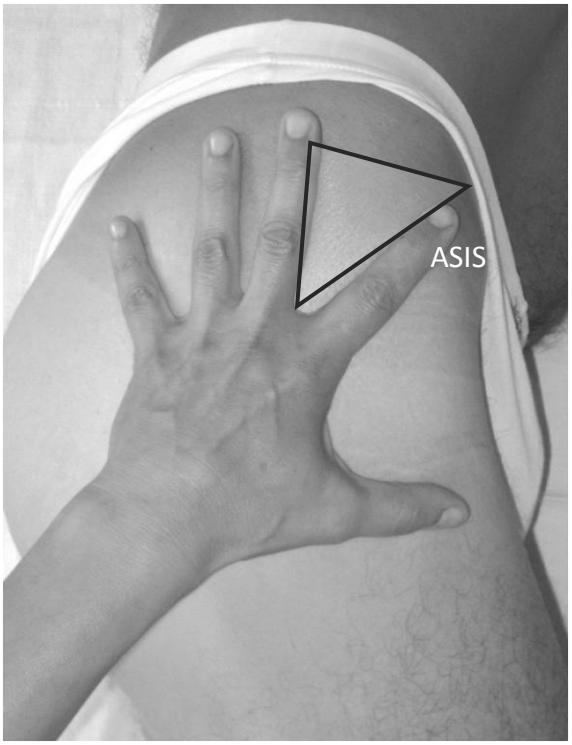


Figure 3.10. Location of the 'von Hochstetter triangle', index finger on the ASIS, middle finger on iliac crest, ball of the thumb on greater trochanter (Schuenke et al 2007, pp. 494/495, Mensdorf 1999).

3.4.4 A splint to secure leg rotation

The experience of Setup One indicated the need to secure leg rotation by a device because some subjects needed activity to keep the leg in neutral rotation and could not relax fully. In Setup Two, a splint around the calf that minimized leg rotation was used. The shank with the splint was positioned on a sliding board (Figure 3.11).

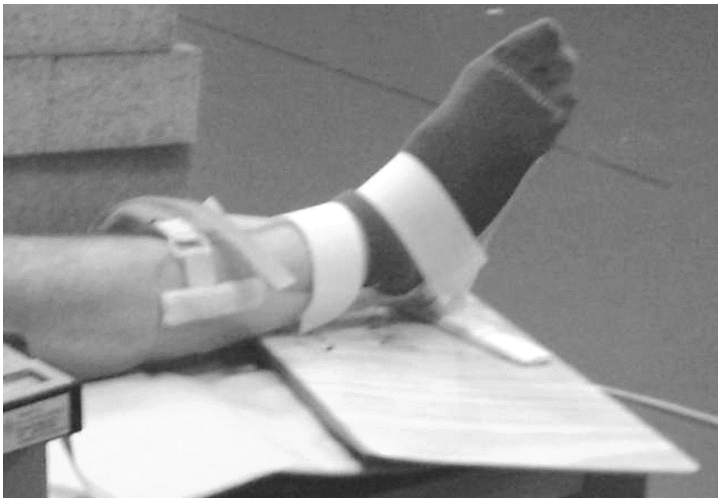


Figure 3.11. Leg rotation was secured by a splint around the calf, which was positioned on a sliding board.

3.4.5 Procedure

Isometric right hip abduction in a neutral joint position was performed in supine against a dynamometer. This was fixed to the right side of the plinth and was adjustable in height and position. Subjects were positioned parallel to the side of the plinth and with contact of the distal, lateral thigh to the dynamometer. On the left side of the pelvis, a plate was fixed for pelvic stabilization. Following several testing abductions, the level of maximal isometric abduction force (MVIC) was determined in three ramp activations. Providing visual feedback of the actual exerted force and the required target force levels, three repetitions of 20% and 40% and two repetitions of 60% MVIC were recorded without designating activity speed. Finally, three repetitions of 60% MVIC in maximal activation speed were requested.

A second task in weight-bearing had been planned. Due to pain by the fine-wire electrodes in one-legged stance, data collection in weight-bearing had to be stopped.

3.4.6 Instrumentation

3.4.6.1 Dynamometry

Measurements of the exerted torque were collected at 2000 Hz using a Mecmesin dynamometer (AFG-500N, Slinfold, UK). Signals were low-pass filtered with a zero lag 4th order Butterworth filter at 40 Hz.

3.4.6.2 Fine-wire EMG

Fine-wire EMG was recorded on an Octopus AMT-8 EMG system (Bortec Electronics Inc., Calgary, Canada) at a sampling rate of 2000 Hz, input impedance 10 GOhm, common mode rejection of 115 dB at 50 Hz. Two pairs of custom prepared (Hodges 2006), differential, Teflon-coated fine-wire electrodes (California Wire Company, Grover Beach, USA) 0.075 mm diameter were inserted in a sterile procedure under US guidance obliquely into the deep gluteus medius and minimus muscles. Coating at the electrode ends had been removed and hooks formed to ascertain the final position after insertion. For insertion, 7 cm long 23G needles (Terumo Cooperation, Tokyo, Japan) were used. Beside US control, piercing of the gluteus minimus aponeurosis provided a definite event to facilitate distinction of the two insertion levels. Free electrode ends were connected to amplifiers and secured. A surface ground electrode was positioned on the contralateral ASIS. Following a test of electrode function, the insertion area was covered against US gel with gauze and a transparent adhesive film (OPSite, Smith & Nephew, London, UK). Raw signals were pre-amplified 500 times and amplified to display amplitude of ca. 2 mV with ramp activation. Signals were demeaned, zero-lag band-pass filtered (10 - 900 Hz) and full-wave rectified. EMG and torque measurements were recorded and displayed by Vicon Nexus software (Vicon Motion Systems Ltd, Oxford, UK).

3.4.6.3 M-mode ultrasound

Gluteus medius and minimus activity was scanned using a Xario XG, SSA-680A (Toshiba Pty Ltd, North Ryde, NSW, Australia), linear probe PLT-704SBT, 7.5 MHz, 3.8 cm footprint. The US transducer was attached to a custom-prepared, wedge-shaped support (Figures 3.13 and 3.14), positioned over the fine-wire electrodes and fixed by an elastic belt. M-mode was set in a custom- programmed application with strong contrast, high frame rate and highest sweep speed. The temporal resolution of the M-mode trace was 2.2 ms.

3.4.6.4 Transducer fixation

Transducer fixation in a foam block occupies space that is needed for EMG electrodes. The foam block does not allow for visual control of the exact transducer position. A custom-made transducer fixation (Figures 3.12 and 3.13) was developed that overcame these disadvantages. The new device consisted of a triangular shaped support made from thermoelastic splint material with a Velcro surface. The transducer was wrapped with Velcro, which allowed for firm attachment on the support.

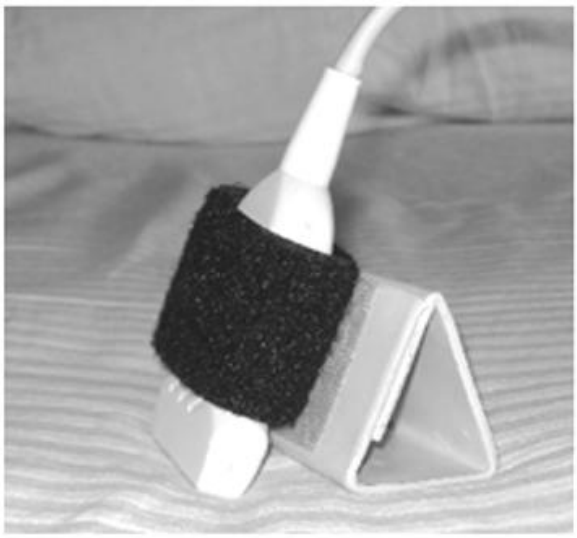


Figure 3.12. Custom-made transducer support from thermoelastic material and Velcro.

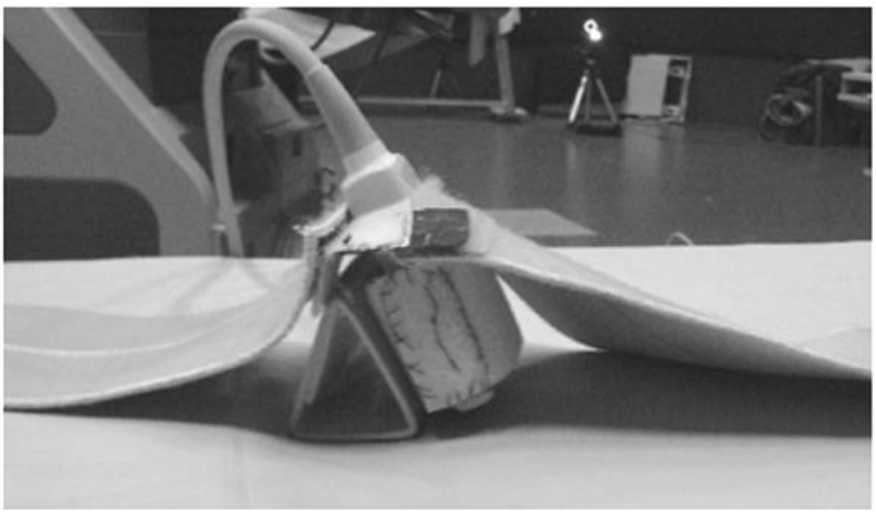


Figure 3.13. Transducer on custom-made support with belt for fixation.

3.4.6.5 Synchronization

An electrical signal triggered by a footswitch was split to go synchronously into a separate channel of the Vicon data collecting system and into the ECG entrance of the US system to produce a visual signal on the bottom of the M-mode trace. A potential delay in the ECG channel due to system-internal processing (Walker et al 2002) (Chapter Six, section 6.2) had been tested and found to be 2ms.

3.4.7 Questions to be answered from the data collected in Setup Two

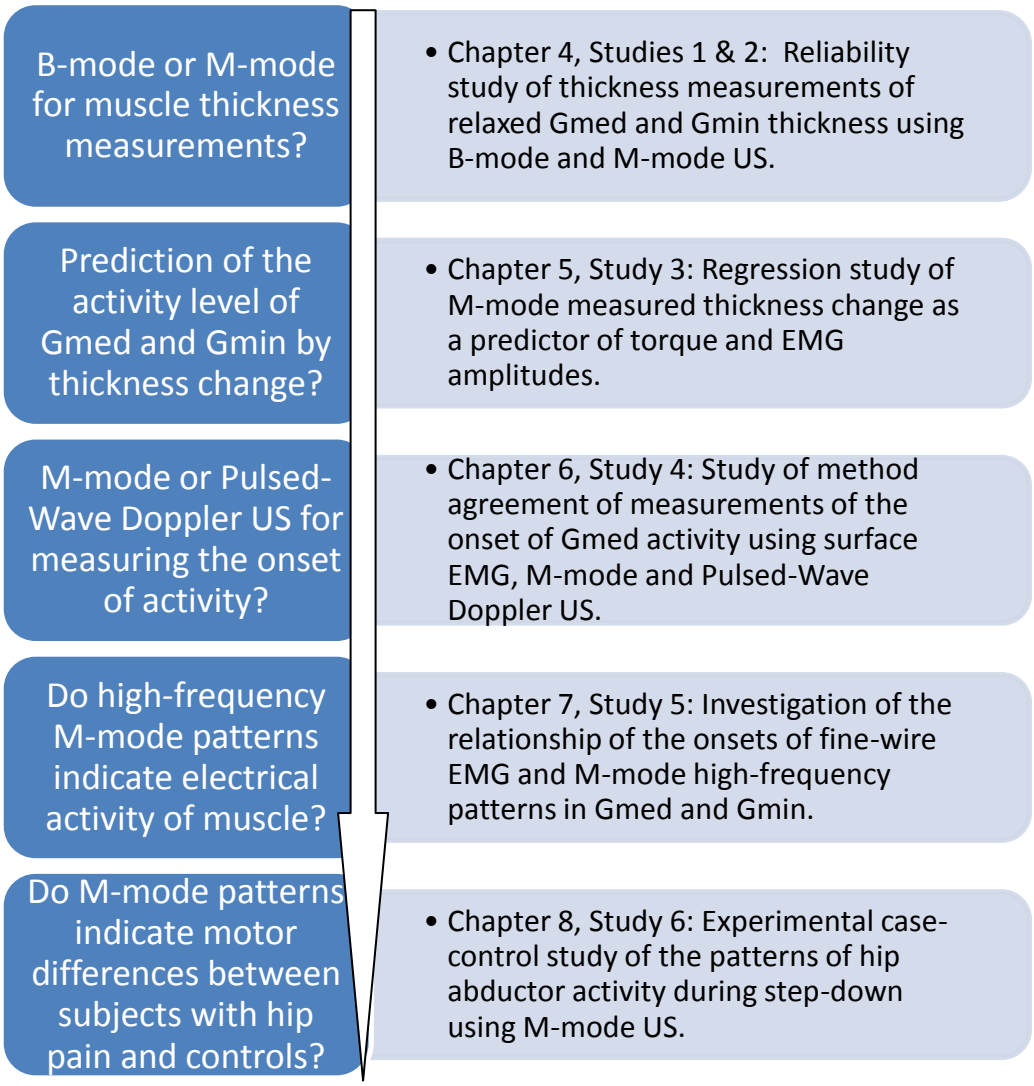
Is the level of electrical excitation of gluteus minimus predictable by M-mode measurements of thickness change?

The objective to estimate the prediction of the force level of isometric hip abduction, as measured by dynamometry and fine-wire EMG amplitude, from the change of gluteus minimus *thickness* is examined in Chapter Five.

Is electrical excitation of the deep gluteus medius and minimus muscles indicated by M-mode US motion measurements?

The objective to determine the relationship between measurements of activity *onset* by fine-wire EMG and M-mode US in the deep gluteus medius and minimus muscles is investigated in Chapter Seven.

Chapter Four initiates the investigation of study questions by analysis of data collected in Setup One. The following figure serves for orientation in the progress of the research process.


B-mode or M-mode for muscle thickness measurements?
• Chapter 4, Studies 1 & 2: Reliability study of thickness measurements of relaxed Gmed and Gmin thickness using B-mode and M-mode US.
Prediction of the activity level of Gmed and Gmin by thickness change?
• Chapter 5, Study 3: Regression study of M-mode measured thickness change as a predictor of torque and EMG amplitudes.
M-mode or Pulsed-Wave Doppler US for measuring the onset of activity?
• Chapter 6, Study 4: Study of method agreement of measurements of the onset of Gmed activity using surface EMG, M-mode and Pulsed-Wave Doppler US.
Do high-frequency M-mode patterns indicate electrical activity of muscle?
• Chapter 7, Study 5: Investigation of the relationship of the onsets of fine-wire EMG and M-mode high-frequency patterns in Gmed and Gmin.
Do M-mode patterns indicate motor differences between subjects with hip pain and controls?
• Chapter 8, Study 6: Experimental case-control study of the patterns of hip abductor activity during step-down using M-mode US.

4 Chapter Four: M-Mode in the Longitudinal Scanning Plane Enables Reliable Thickness Measurements and the Identification of Muscle Relaxation

Overview of Chapter Four

Studies One and Two in this chapter were preparatory and were undertaken to guide the methodological decisions for Study Three in which the prediction of abductor activation levels from muscle thickness change is investigated.

- Section 4.1 presents Study One in which gluteus medius and minimus thickness measurements by B-mode and M-mode US were compared for their reliability and for the additional information provided by each mode.
- Section 4.2 reports a small study, Study Two, in which longitudinal and transverse scanning of hip abductor activity by M-mode US was compared for the consistency of the scanning angle.
- Section 4.3 summarizes the main conclusions of Chapter Four for the application of US measurements of hip abductor muscle thickness.

4.1 Study One: Thickness measurements of gluteus medius and minimus: B-mode or M-mode ultrasound?

4.1.1 Introduction

The anatomical differences between gluteus medius and gluteus minimus which are presented in section 2.6 of the literature review suggest functional differences between the two abductors. Functional differences may be reflected by a differential force contribution or timing of activity. When US imaging is used for assessing muscle activity, the measurement of muscle thickness is the most widely reported measure (Chapter Two, section 2.5.1). Muscle thickness has been measured by B-mode US (Ikezoe et al 2011, O'Sullivan et al 2007, Stokes et al 2007, Teyhen et al 2007, Whittaker et al 2007) and also by M-mode (Mannion et al 2008b, Bunce et al 2004, McMeeken et al 2004, Kidd et al 2002). B-mode offers the advantage of anatomic orientation in a two-dimensional image, but the measurement location and angle have to be set manually in each image. In M-mode, the measurement location and angle is fixed once the M-mode beam is positioned, which may be advantageous for standardization and comparison. M-mode offers the additional advantage of high temporal resolution for monitoring muscle motion, allowing not only for thickness but also for motion measurements (Vasseljen et al 2009, Vasseljen et al 2006). Mc Meeken et al compared thickness measurements of the transversus abdominis muscle by B- and M-mode US and reported no relevant difference between the modes (2004). It is unclear whether these results are reproducible on the hip abductors and whether the additional information provided by each mode supports a decision which mode should be chosen.

Reliable thickness measurements at the lateral pelvis may be particularly complicated because a non-perpendicular scanning approach is required. The oblique iliac bone and muscle fascias are not parallel with the skin surface (Grimaldi 2011, MRI section). The reliability of gluteus medius and minimus thickness measurements depends on a consistent scanning angle (Whittaker et al 2009, Klimstra et al 2007).

Muscle thickness can be measured on the relaxed and the activated muscles. Study One compares B-mode and M-mode baseline measurements of the relaxed abductor muscles. Measurements of relaxed muscle thickness serve as baseline for the measurement of the thickness change during activity. The stability of baseline

measurements indicates the precision of thickness measurements to detect small changes of muscle thickness by activity. The objective of Study One was to compare baseline thickness measurements of the relaxed gluteus medius and minimus muscles with regard to the intra-tester reliability, within and between days, and to the additional information provided by each mode.

4.1.2 Methods

The methods of data collection are described in detail in Chapter Three, sections 3.1 and 3.3 and briefly here.

4.1.2.1 Subjects

Asymptomatic volunteers were recruited in the School of Physiotherapy, Curtin University. Exclusion criteria were hip pain, history of hip pathology or general musculoskeletal disease and BMI > 32 kg/m^2. Ethical approval was obtained (PT0107/2007). Subjects provided informed consent.

4.1.2.2 Procedure

Baseline measurements of relaxed muscle thickness should be repeatable in spite of intermittent phases of muscle activity. On two testing occasions within one week, subjects performed three maximal voluntary isometric contractions (MVIC) of right hip abduction, in supine, sustained for four seconds per contraction, with relaxation intervals of 60 seconds. Two MVIC trials and relaxation breaks were recorded by B-mode US imaging and one by M-mode US of the gluteus medius and minimus muscles (Figure 3.9).

4.1.2.3 US imaging

The right hip abductors were imaged with an Antares 4.0 system (Siemens Medical Solutions, USA), a linear probe (VFX9-4) set to 9 Mhz, 38 mm footprint. The transducer was housed in a foam block excavated at a 20° angle which was fixed to the pelvis. The scanning location on the right hip was determined on the lower half of a reference line that connected the tip of the greater trochanter with the anterior quarter of a line connecting the anterior and posterior superior iliac spines (Chapter Three, Figure 3.1).

In B-mode US, a custom application was programmed to delineate high contrast and low persistence to avoid blurred images by motion. The M-mode settings included a small B-mode image for anatomic orientation on top of the M-mode trace which represented 1.4 seconds of tissue motion at highest sweep speed.

US recordings of the complete MVIC testing procedure were captured at 25 frames per second onto video (Panasonic digital video camera, NV-MX 500A, Secaucus, USA).

Gluteus medius relaxation was controlled by EMG (Octopus AMT-8 EMG system, Bortec Electronics Inc., Calgary, Canada). Following skin preparation, two surface electrodes (3M Australia, Pymble, NSW) were attached on the reference line cranial to the foam block and a ground electrode on the lateral ribs. The electrodes were positioned over the same gluteus medius part as the US transducer. US and EMG data were synchronized using a PEAK synchronization unit (Event Synchronization Unit, PEAK Performance Technologies Inc., Centennial, CO, USA) which produced a time stamp on the concurrent video frame when EMG recording was started.

Muscle relaxation was confirmed by baseline gluteus medius activity on the EMG and no visible motion in the US clip. Two B-mode and two M-mode frames of the relaxed gluteus medius and minimus muscles were exported as Tiff files from the videos of the stable relaxation phases before, between and after the MVIC tests (iMovie, version 5.0.2).

4.1.2.4 US measurements

Thickness of the relaxed gluteus medius and minimus muscles was measured off-line on pairs of enlarged (150%) images of the same subject using ImageJ software (version 1.40; rsb.info.nih.gov/ij/). Images were displayed in pairs to facilitate the recognition of muscle fascia. The examiner was blinded to measurements from the other mode or the other occasion. Measurements were taken at the location of the thickest muscle bulk and from the inner edges of the fascias (Whittaker 2007, p. 99).

The B-mode measurements were taken perpendicular to the muscle fascias. M-mode measurements were taken aligned to the US transducer's long axis (Figure 4.9), as the angle of the M-mode beam could not be steered to cross the muscle perpendicularly. Consequently, the angle of the B-mode and the M-mode measurements may differ. In order to compare the absolute thickness values between

US modes, additional, B-mode measurements aligned to the US transducer were taken (Figure 4.1).

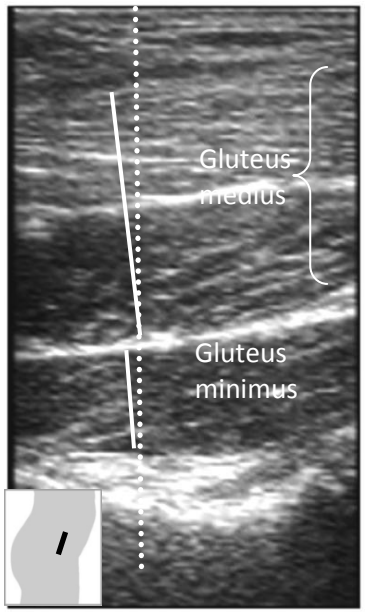


Figure 4.1. B-mode measurements of muscle thickness were taken along the continuous white lines. Additional, measurements aligned to the US transducer were taken along the dotted line for comparison with the M-mode measurement.

Depending on the arrangement in the scanning setup, the preset M-mode measurement line may introduce measurement variation by differences in the angle of the M-mode beam towards the muscle fascia. To document the angle variation, the angles between the aponeurosis, the periosteum and the vertical measurement line were measured (Figure 4.2).

Muscle thickness was measured a second time one week later on all images of the first occasion (Day 1) to estimate within-day intra-tester reliability. The whole imaging procedure was repeated within four days (Day 2) to assess between-days reliability.

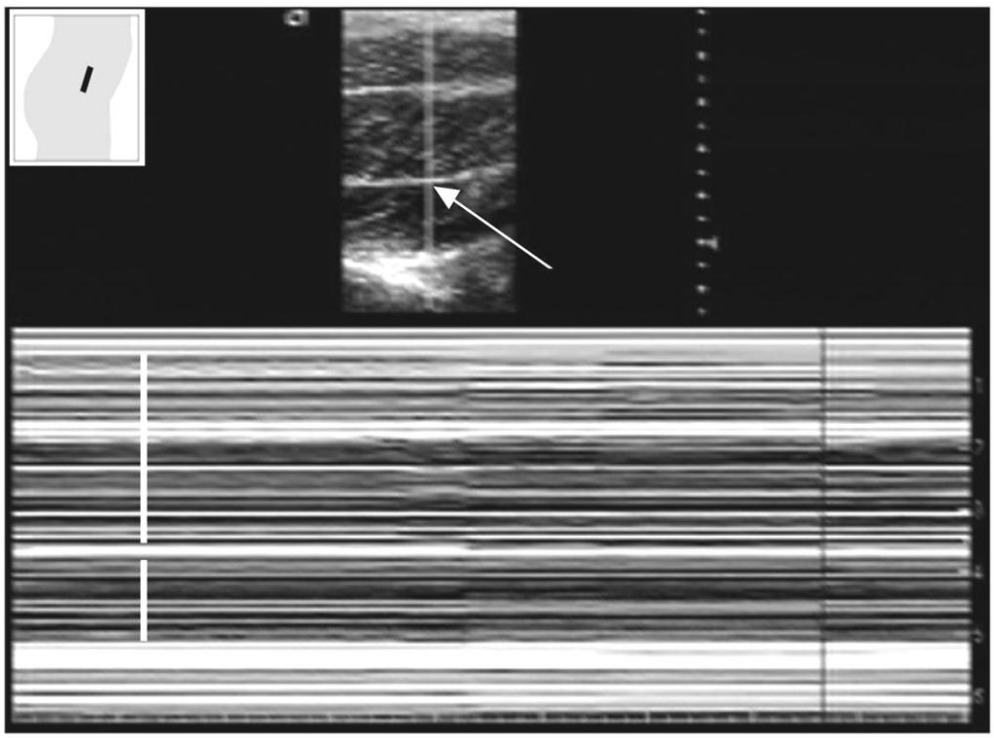


Figure 4.2. M-mode measurements of muscle thickness were taken along the vertical, white line. The angles of the M-mode measurement line towards the aponeurosis (arrow) and the periosteum were measured to document angle variation.

4.1.2.5 Statistics

Intra-tester reliability of all measurements of Day 1 and between-days reliability of the measurements of Day 1 and Day 2 were estimated by intra-class correlation coefficients $ICC_{3,1}$ for single and $ICC_{3,k}$ for averaged measures. Means, standard deviations (SD), standard errors of measurement (SEM) of the difference between Day 1 and 2 measurements and minimal detectable change (MDC, MDC = SEM*√2*1.96, (Donoghue et al 2009) were determined. The MDC is a statistical estimate for the change which can be considered with 95% certainty as 'true' and not measurement error (Donoghue et al 2009). The MDC as percentage of total muscle thickness indicates how much percent of thickness change needs to happen to be 95% confident that the muscle is active. The variation of the angle between the M-mode beam, gluteus minimus aponeurosis and periosteum was described by mean and SD. Differences between B-mode and M-mode measurements of muscle thickness were examined by paired *t*-tests. Significance was set to $\alpha = 0.05$. If not specified otherwise, results are reported as mean (SD).

4.1.3 Results

Sixteen volunteers participated, ten females aged 30.1 (10.6) years with BMI of 22.6 (4.2) kg/m^2 and six males aged 28.8 (6.2) years with BMI of 22.2 (2.6) kg/m^2. Three subjects were overweight with BMIs of 26.9, 28.7 and 30.9 kg/m^2. Six females and five males also participated in testing on Day 2. On each day, two B-mode and two M-mode images of the relaxed abductor muscles were recorded, resulting in 108 images.

High intra-tester reliability within-day (Table 4.1) and slightly lower between-days reliability (Table 4.2) were indicated by ICCs in excess of 0.78, slightly higher for averaged than for single thickness measurements.

The MDC for measurements of the same occasion was 6 - 9% of mean gluteus medius thickness and 9 - 16% of mean gluteus minimus thickness. The MDC for measurements between days ranged from 10 - 13% of gluteus medius and 15 - 18% of gluteus minimus thickness.

The angle of the M-mode measurement line towards the aponeurosis was 82.3° (8.8°), and towards the periosteum 90.0° (10.0°) (Figure 4.2).

Table 4.1. Intra-tester reliability within-day of relaxed gluteus medius and minimus thickness measurements in B- and M-mode ultrasound, intraclass correlation coefficients (ICC), confidence intervals (CI), standard error of measurement (SEM) and minimal detectable change (MDC, $=\mathrm{SEM} \times \sqrt{2} \times 1.96$).

n = 16	**B-mode gluteus medius**	**B-mode gluteus minimus**	**M-mode gluteus medius**	**M-mode gluteus minimus**
Thickness (SD) mm	23.8 (2.8)	16.1 (2.3)	23.3 (2.1)	15.5 (2.1)
Averaged measures, ICC $_{3,k}$ (CI)	0.983 (0.951 - 0.994)	0.973 (0.922 - 0.991)	0.966 (0.903 - 0.988)	0.914 (0.754 - 0.970)
Difference ± SD, mm	0.0 ± 0.8	0.2 ± 0.8	-0.1 ± 0.8	-0.5 ± 1.2
SEM, mm	0.5	0.5	0.6	0.8
MDC, mm	1.5	1.5	1.5	2.3
MDC % of thickness	6.2%	9.3%	6.6%	15.1%
Single measures, ICC $_{3,1}$ (CI)	0.954 (0.907 - 0.977)	0.932 (0.866 - 0.966)	0.885 (0.778 – 0.942)	0.802 (0.633 - 0.898)
Difference ± SD, mm	0.0 ± 0.9	0.2 ± 0.9	-0.1 ± 1.1	-0.5 ± 1.3
SEM, mm	0.6	0.6	0.7	1.0
MDC, mm	1.7	1.7	2.1	2.6
MDC % of thickness	7.3%	10.6%	8.9%	17.0%

Table 4.2. Intra-tester reliability between-days of relaxed gluteus medius and minimus thickness measurements in B- and M-mode ultrasound, intraclass correlation coefficients (ICC), confidence intervals (CI), standard error of measurement (SEM) and minimal detectable change (MDC).

n = 11	**B-mode gluteus medius**	**B-mode gluteus minimus**	**M-mode gluteus medius**	**M-mode gluteus minimus**
Thickness (SD) mm	23.2 (2.8)	16.6 (2.2)	23.5 (3.5)	17.4 (2.5)
Averaged measures, ICC $_{3,k}$ (CI)	0.948 (0.808 - 0.986)	0.888 (0.584 - 0.970)	0.938 (0.626 - 0.967)	0.897 (0.618 - 0.972)
Difference ± SD, mm	0.3 ± 1.5	0.5 ± 1.6	-0.3 ± 1.5	0.6 ± 1.5
SEM, mm	1.0	1.1	1.1	1.1
MDC, mm	2.4	2.5	3.0	3.0
MDC % of thickness	10.4%	14.9%	12.6%	17.3%
Single measures, ICC $_{3,1}$ (CI)	0.884 (0.742 - 0.950)	0.788 (0.556 – 0.906)	0.869 (0.711 - 0.943)	0.803 (0.584 – 0.913)
Difference ± SD, mm	0.3 ± 1.6	0.5 ± 1.6	-0.3 ± 1.6	0.6 ± 1.6
SEM, mm	1.1	1.1	1.1	1.1
MDC, mm	2.5	2.6	3.1	3.0
MDC % of thickness	10.7%	15.4%	13.1%	17.5%

4.1.4 Discussion

Hip abductor thickness can be measured by B-mode or M-mode US; in both modes good to excellent measurement reliability was demonstrated. The variation of baseline measurements of relaxed muscle thickness was slightly less in B-mode, resulting in a 0.4-6.4% higher sensitivity to detect activity. M-mode revealed isolated gluteus minimus motion not shared by gluteus medius. This observation indicates the unique property of M-mode US to enable a visual assessment of relaxation in the deep gluteus minimus muscle..

Study One supports the high reliability of gluteus medius and minimus thickness measurements reported by Ikezoe et al (2011). The ICCs of the intra-tester reliability of gluteus medius and minimus thickness measurements within-day (Costa et al 2009, Hebert et al 2009) and between-days (Koppenhaver et al 2009a, Rankin et al 2006) were in the range reported in the literature on trunk muscle measurements. Marginally lower ICCs for M-mode compared to B-mode measurements were indicated, a result in concordance with McMeeken et al (2004). The absolute measurement error of within-

day reliability, as expressed by the SEM, was in this study comparable to measurements on the abdominals by Mannion et al (2008b) and by Critchley and Coutts (2002), but higher than values reported by Hides et al (2007) and Teyhen et al (2005). Absolute reliability between days, as indicated by SEM, was in this study comparable to the results reported by Kidd et al (2002) and slightly worse than those by Hides et al (2007). Considering the larger depth of muscles, the inclusion of overweight subjects and the difficult, oblique scanning approach, the documented measurement reliability indicates a reasonable measurement quality.

In spite of similar SEMs in gluteus medius and minimus, the percentage of MDC is higher in the thinner gluteus minimus which indicates the principal difficulty to detect small changes of thickness in thin muscles (Teyhen et al 2011, Koppenhaver et al 2009a). The precision of US measures of muscle thickness is not good enough to detect small thickness changes. Three points of importance for the precision of muscle thickness measurements are discussed in the following sections, the identification of the muscle fascias, the consistency of the angle of the sound beam towards the muscle fascias and the control of muscle relaxation.

4.1.4.1 Recognition of muscle fascias

The B-mode image is a reflection of an anatomical cross section, compromised by speckle and reflective phenomena. The two-dimensional B-mode image facilitates the recognition of muscle fascias. The identification of structures in a B-mode image is based on their spatial arrangement, their shape, the continuity and the brightness of lines. The M-mode trace, which is based on a single sound beam, offers fewer characteristics to identify an anatomic structure. Every reflective interface in the sound beam produces a continuous line in the temporal dimension of the M-mode trace, not only fascias but also each fascicle. In M-mode, beside the accompanying B-mode image, the thickness, the brightness and the sequence of lines are the prominent characteristics for identification. Line brightness can be deceptive due to the anisotrophic nature of muscle tissue. Anisotrophy is a visual phenomenon in US imaging of highly reflective tissue. Anisothrophy comprises marked changes in brightness by small changes of the scanning angle (Van Holsbeeck and Introcaso 2001, pp. 17/18). The effect of anisotrophy on thickness measurements may be larger in M-mode because anatomic recognition relies more strongly on the brightness of lines.

The study experience indicated that fascia recognition is facilitated by dynamic observations of muscle and by grouping images of the same subject. The comparison of two or more similar but not identical images helps to recognize a typical pattern of line sequence and therefore the fascias. Mc Meeken et al (2004) reported less difference between B-mode and M-mode thickness measurements than found in this study. A possible reason is that McMeeeken et al used a set-up in which the B-mode image was side by side with an M-mode trace of reduced length, whereas in this study a small B-mode image was on top of a full-length M-mode trace. The full-length M-mode trace reflects a longer 'history' of muscle movement and facilitates the recognition of relaxation and sustained activation. The B-mode image aside of the M-mode trace allows for easier fascia recognition.

4.1.4.2 The angle of thickness measurements

Measurements of muscle thickness should be taken perpendicular to the fascias (Teyhen et al 2007). The manual setting of the B-mode measurement line enables perpendicular thickness measurements regardless of the muscle's orientation in the image. In M-mode, the sound beam, along which thickness is measured, is aligned with the transducer. M-mode thickness measurements perpendicular to the muscle fascias require a horizontal arrangement of the muscle itself in the US image, which may be difficult to achieve. The angles documented here of the M-mode sound beam towards the muscle borders indicated that measurements aligned with the fixed sound beam increased measurement variation due to variation of the muscle arrangement in the image. The documented deviations were within 10°. According to studies by Whittaker et al (2009) and Klimstra et al (2007), angle deviations within 10° should not affect thickness measurements significantly. Nevertheless, angle variation increased measurement variation and decreased the precision of the M-mode measurements of 'true' muscle thickness. This disadvantage of M-mode for thickness measurements can be reduced by arranging the muscles perpendicular to the transducer in the image setup. A technical solution is provided by a beam-steering option, as utilised in Doppler US and available in some US systems. M-mode beam steering would reduce the variation of M-mode thickness measurements.

4.1.4.3 Observation of muscle motion and control of relaxation

According to Hodges et al (2003) most thickening of muscle during activity occurs in low levels of torque. Therefore, control of muscle relaxation is crucial for establishing a constant baseline value of relaxed muscle thickness. Muscle activation is accompanied by the motion of muscle tissue (Vasseljen et al 2009, Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006). M-mode enables the observation of muscle motion by delineation of the displacement of reflective interfaces. M-mode is highly sensitive to motion. Figure 4.3 demonstrates a repetitive, regular pattern of minimal gluteus minimus motion elicited by a pulsating vessel (verified by B-mode and Doppler US). Also, the relaxed state before activity can be distinguished visually from sustained activity. The onset of activity can be visually recognized (Figure 4.4). Clinically, the high sensitivity of M-mode for motion is used to detect fasciculations and fibrillations (Dengler 2009, Pillen et al 2009, Wenzel et al 1998, Reimers et al 1996).

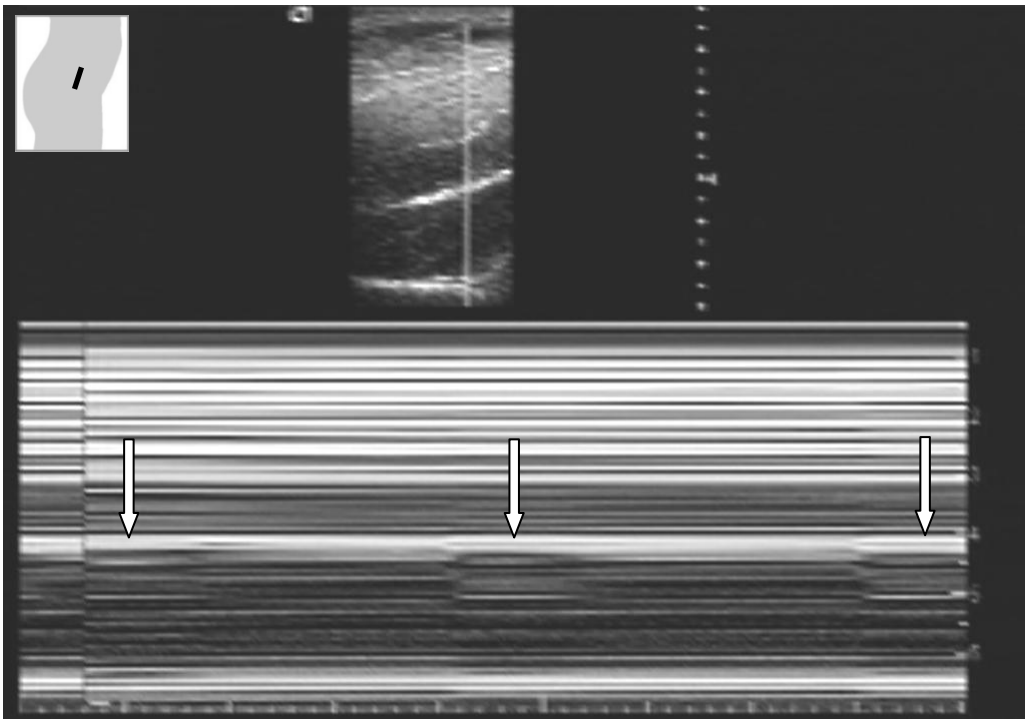


Figure 4.3. Regular appearance of slight gluteus minimus motion (arrows) due to a pulsating vessel.

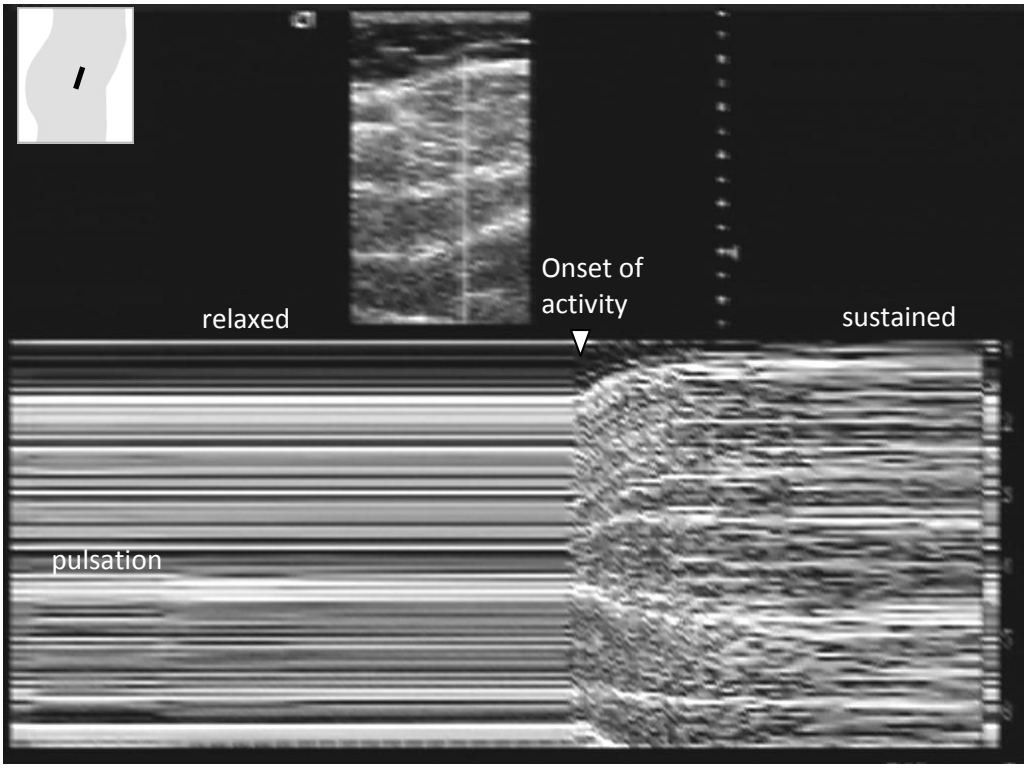


Figure 4.4. The relaxed state before activity can be distinguished visually from sustained activity (MVIC trial, isometric abduction, verified by surface EMG and force trace).

During Study One, isolated and nearly isolated gluteus minimus motion was observed (Figures 4.5 and 4.6), possibly an indication of selective, low level activity of gluteus minimus. Selective gluteus minimus activity cannot be controlled by surface EMG and is difficult to observe in B-mode US, but it can be recognized easily in M-mode US. The interpretation of muscle motion may be ambiguous because motion need not indicate muscle excitation (Vasseljen et al 2006). However, muscle relaxation is unambiguously indicated by non-moving, horizontal M-mode lines of muscle tissue. The easy recognition of relaxation in deep muscles is an important advantage of M-mode US.

Table 4.3 summarizes the main points of the comparison of B-mode and M-mode US for measurements of muscle thickness.

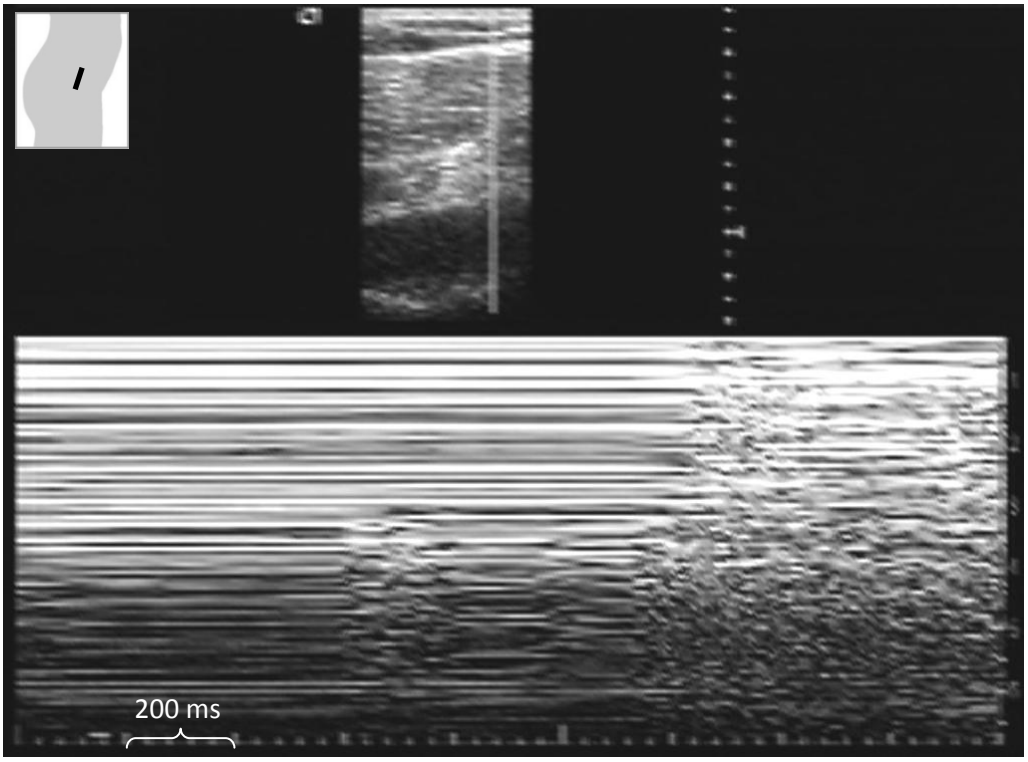


Figure 4.5. Isolated gluteus minimus motion that preceded a 20% MVIC isometric abduction activity of gluteus medius by 640 ms.

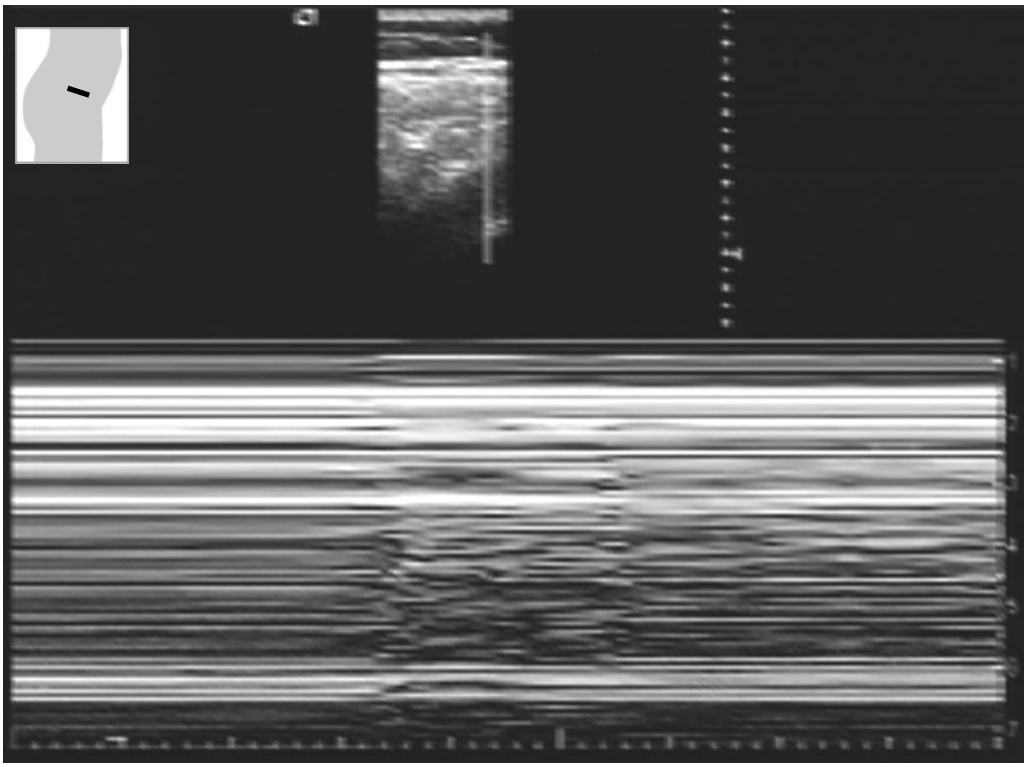


Figure 4.6. Marked motion in gluteus minimus while gluteus medius is fairly relaxed.

Table 4.3. Summary of the main advantages and disadvantages of B-mode and M-mode US for measuring muscle thickness.

	Within-day reliability	Between-days reliability	Muscle border recognition	Consistency of measurement angle	Control of muscle relaxation
B-mode, SEM mm	ICC > 0.97 0.5	ICC > 0.88 1.0 – 1.1	Facilitated by 2-dimensional image	Can be achieved manually	No image of motion
M-mode, SEM mm	ICC > 0.91 0.6 -0.8	ICC > 0.89 1.1	Needs comparison to B-mode	Has to be achieved initially in the setup	Trace of motion over time, highly sensitive to motion

4.1.5 Study limitations

Study One investigated the basic question for the US mode that needed to be decided before the investigation of thickness change. The focus of this study is limited for the clinician who may ask for the inclusion of thickness measurements of activity, an investigation that is presented in the following chapter. In spite of the limitation to measurements of the relaxed muscles, the main characteristics of B-mode and M-mode US thickness measurements are reflected in this study. No comparable comparison between B- and M-mode US has been reported in the literature.

From a clinical perspective, a reliability study in which the US transducer has been fixed by a device may be questioned, because fixation devices have to be custom-made and their attachment is time-consuming. Experienced professionals who trained maintaining the scanning angle while interacting with the subject may not need transducer fixation. To date, US imaging skills are not the focus of physiotherapy training and rather a supplementary assessment. Therefore, a fixation device is recommended to control the scanning angle and transducer pressure in repeated measurements. The precise reproduction of the scanning angle is critical for thickness measurements (Whittaker et al 2009, Klimstra et al 2007). The oblique scanning angle required for gluteus medius and minimus is particularly difficult to reproduce, even more in a rehabilitation setup that includes active tasks and limb motion. A further argument for a fixation device is the control of the pressure of the US transducer. Pressure of the transducer may affect muscle thickness, in particular in subjects with a thin layer of adipose tissue. The foam block provides a uniform pressure as it distributes the pressure by fixation over a larger surface than the footprint of the US transducer.

The foam block, not the transducer itself was fixed to the pelvis, so that no specific pressure was exerted on the transducer handle. The foam block was of medium density. Pressure on the foam block deformed the foam before being transmitted to the muscles.

As documented in Figures 4.2 to 4.6, the image quality provided by video recording is limited. A digital recording technique would enhance image quality. From a statistical point of view, the sample sizes of sixteen for within-day and of eleven for between-days reliability may be judged insufficient (Walter et al 1998), although reliability studies with comparable sample sizes are common in well-recognized journals in the medical imaging field (E Lima et al 2012, Dudley-Javoroski et al 2010, Alshami et al 2009). An important limitation of this study was that gluteus minimus relaxation was not controlled by fine-wire EMG. Ethically, for the control of relaxation alone, the invasive procedure cannot be justified. The observations in this study suggest that M-mode is sufficient to control for relaxation because the interpretation of non-moving muscle being relaxed is unambiguous.

4.1.6 Conclusions

Intra-tester reliability of B-mode and M-mode thickness measurements of gluteus medius and minimus was high, with a slight preference for B-mode. The higher reliability of B-mode is of particular relevance in thickness measurements of thin muscles. B-mode and M-mode thickness measurements did not differ in between-days reliability. M-mode US enables the control of relaxation in deep muscles, a unique advantage for establishing 'true' baseline measurements of muscle relaxation.

It was decided to continue the research on thickness measurements of hip abductor activity using M-mode, firstly due to the measurements of thickness change requiring the control of relaxation, and secondly because onsets of muscle activity were to be examined, an objective which can be studied by M-mode but not by B-mode.

4.1.7 Acknowledgements

The support by Siemens Medical Systems, Western Australia for the loan of equipment and programming of the application and by Paul Davey for technical advice and help is thankfully acknowledged.

Having identified the US mode that will be used for the investigation of thickness changes associated to gluteus medius and minimus activity, a decision on the scanning plane for gluteus medius and minimus thickness measurements is required. The following study was undertaken to inform this basic decision.

4.2 Study Two: Thickness measurements of hip abductor activity by M-mode ultrasound imaging: longitudinal or transverse scanning plane?

4.2.1 Background

Two main scanning planes are used in US imaging. The longitudinal plane is orientated parallel to the longitudinal axis of the target and the transverse plane is perpendicular to the longitudinal plane. The sparse literature which reports comparisons of US thickness measurements in both planes documents conflicting results; no significant difference between the planes has been found when measuring the thickness of the tibial muscle group (McCreesh and Egan 2011), whereas transversely scanned measurements of ulnar nerve diameters were found to be more accurate than longitudinally scanned measurements (Bartels et al 2008). None of these reports examined the changes of muscle thickness during activity.

The first criterion for the choice of a scanning plane may be whether a muscle's cross-section and its changing shape during activity can be demarcated fully in the US field of view. Using a standard 38 mm footprint linear probe, the change of shape of gluteus medius and minimus cannot be fully recognized (Figure 3.3 a). In each plane only a part of the hip abductors can be visualized. It is therefore necessary to consider how the motion that occurs during activity affects thickness measurements in each plane.

Even in isometric activity, muscle tissue moves until the 'slack', the elasticity of the connective tissue, is taken up (Karamanidis et al 2005). In the pennate gluteus medius and minimus muscles, three dimensions of motion can be expected. The fascicles shorten and become more upright (Narici et al 1996, Herbert and Gandevia 1995), the physiological cross-sectional area increases due to fibre thickening and the muscle shortens to take up the tendon's slack (Karamanidis et al 2005). Motion not only

towards the muscles origin but also laterally is possible, as documented for rectus abdominis motion (Brown and McGill 2008).

The skin surface, on which the US transducer is positioned, also changes shape, caused by the bulging muscle. The US transducer moves with the surface, a movement by which the scanning angle towards the muscle may change (Whittaker et al 2010). As the sound beam in M-mode is fixed to the transducer, transducer motion relative to the moving muscle may cause variation in the angle of the sound beam. The objective of this small investigation was to compare the variation of the M-mode sound beam towards the gluteus medius and minimus fascias during isometric activity of the gluteus medius and minimus muscles in the longitudinal and the transverse scanning planes.

4.2.2 Methods

Data of Setup One were analysed. The methodology is presented in detail in Chapter Three, sections 3.1 and 3.3. In short, in supine isometric abduction of the right hip in 80% MVIC against a dynamometer was recorded using M-mode US in the longitudinal and the transverse scanning planes. Activation was sustained for 4 s, followed by a relaxation phase of 45 s. Five cases with good image quality were examined for differences in the angle of the sound beam towards the muscle fascias during the course of an activation. From each subject, an 80% MVIC trial in longitudinal and a second trial in transverse scanning were selected. Five to six frames were cut every 20 ms starting from relaxed through to sustained activity. Using ImageJ (version 1.40; rsb.info.nih.gov/ij/), the angles between the M-mode beam and the inside edges of the upper gluteus medius fascia and the gluteus minimus aponeurosis were measured in the small B-mode images that accompany the M-mode trace. Measurements were taken blinded to previous measurements in the randomly stacked, 200% enlarged images of each subject. The range of angles was documented by mean and standard deviation (SD) for each subject. Differences of sound beam angles were examined by a paired *t*-test. Significance was set to $\alpha = 0.05$.

4.2.3 Results

The angles between the M-mode beam, the upper gluteus medius fascia and the gluteus minimus aponeurosis were measured on 56 images of five subjects. Sound beam angles were different between longitudinal and transverse scanning, *p* .014 for the upper gluteus medius fascia and *p* .000 for the gluteus minimus aponeurosis. Table 4.4. indicates larger angle variation during activity in transverse scanning.

Table 4.4. Angle variation of the M-mode sound beam to gluteus medius and minimus muscle fascias from relaxation to sustained activity, mean and standard deviation (SD).

	Longitudinal, mean (SD)	Difference smallest to largest angle per subject, longitudinal, mean (SD)	Transverse, mean (SD)	Difference smallest to largest angle per subject, transverse, mean (SD)
Upper gluteus medius fascia	86.0 (1.4)°	4.2 (1.5)°	82.7 (5.5)°	6.2 (3.6)°
Gluteus minimus aponeurosis	83.6 (4.8)°	4.7 (2.3)°	64.3 (6.1)°	12.3 (3.9)°

Figures 4.7 and 4.8 illustrate trials of two subjects in longitudinal and in transverse scanning. The first image of each sequence is in relaxation, the middle approximately after 40 ms and the last in sustained activity.

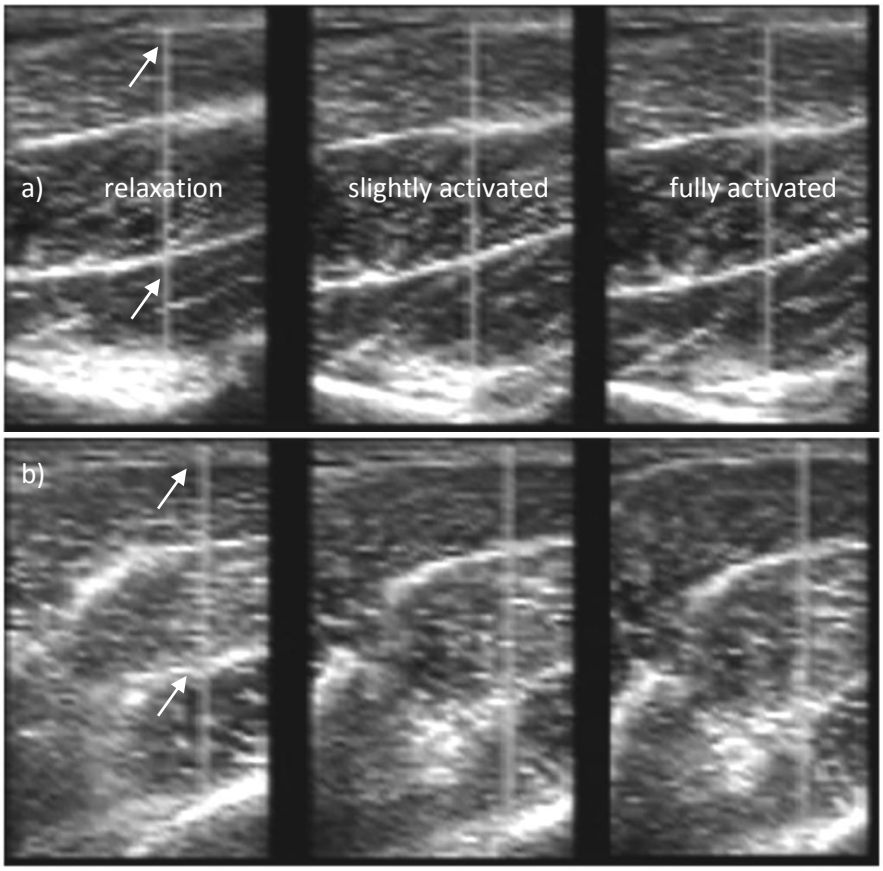


Figure 4.7. Subject 1. Angle variation of the M-mode beam to gluteus medius fascia and gluteus minimus aponeurosis during 80% isometric hip abduction from relaxed to sustained, a) longitudinal scanning, b) transverse scanning.

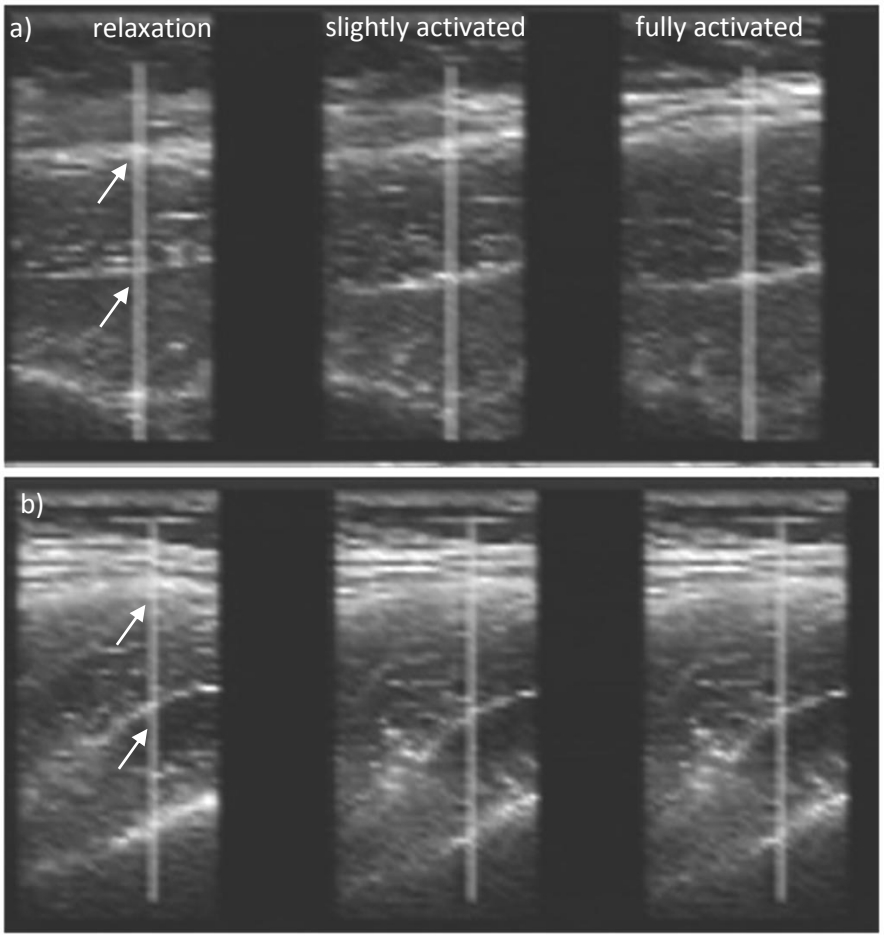


Figure 4.8. Subject 2, Angle variation during 80% isometric hip abduction from relaxed to sustained, a) longitudinal scanning, b) transverse scanning.

4.2.4 Discussion

The results of this small study indicate a higher variation of the angles of the M-mode sound beam towards the gluteus medius and minimus fascias in transverse scanning. The results also document a larger difficulty to position the transducer perpendicular to the hip abductors in a transverse setup.

Research reports indicate that changes of the scanning angle >10° induced significantly different muscle thickness measurements (Whittaker et al 2010, Klimstra et al 2007). In the transverse plane, the angle towards the gluteus minimus aponeurosis changed in mean 12° during an 80% isometric activation, a finding which disqualifies thickness measurements of gluteus minimus in the transverse plane.

To further discuss the effects of motion of the muscle and of the US transducer, a terminology needs to be defined that allows describing the relationship between two three-dimensional bodies which both move. One body is the muscle as part of the scanned subject and the other body is the transducer. Each body can be described in its own, three-dimensional coordinate system (Bergmann et al 2001).

The planes of the human body are termed the frontal (or coronal), sagittal and transverse body planes. Definitions of the two main planes and the main axis of the transducer coordinate system are given in Figure 4.9, the cube representing the transducer. Transducer motions have been defined as *rocking* in the lateral plane, *tilting* in the elevation plane and r*otating* when occurring around the long axis (Ophir et al 1999).

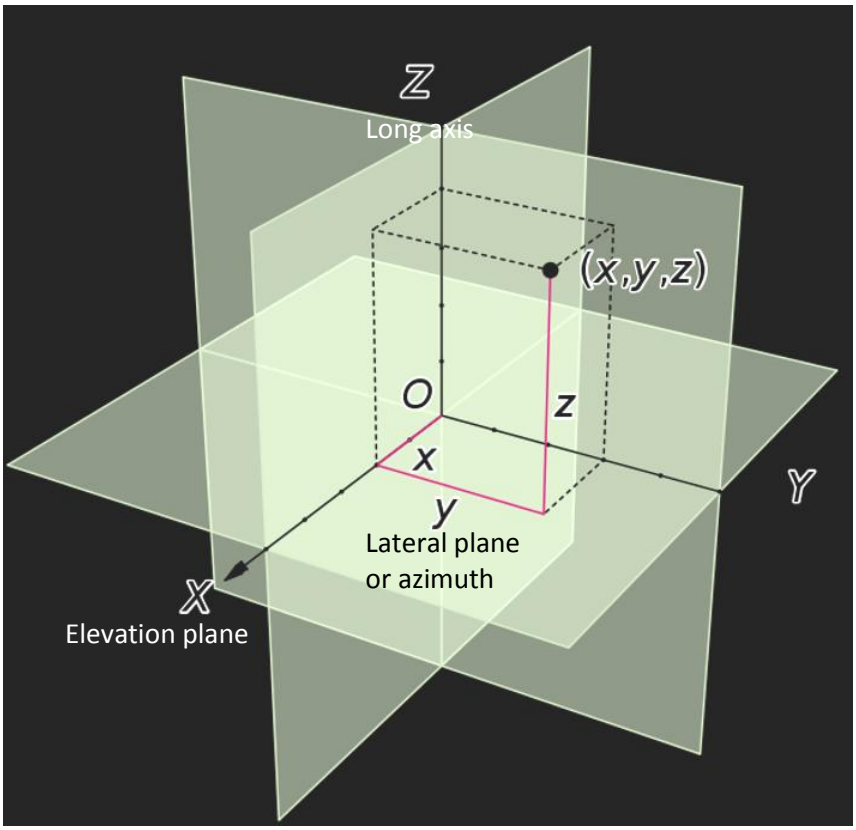


Figure 4.9. Planes and directions of movement of the three-dimensional coordinate system of the ultrasound transducer (represented by the cube).[2]

[2] Free of license illustration by Stolfi J (2009) Three-dimensional coordinate system. http://en.wikipedia.org/wiki/File:Coord_system_CA_0.svg. [Accessed 08/10, 2012].

A rocking motion of the US transducer (= in the lateral plane of the transducer) is detectable in the image and can be quantified by a change of the sound beam angle relative to structures in the image. A tilting motion of the US transducer (= along the elevation plane) changes the scanned tissue section and cannot be recognised in the US image. Tilting motion of the transducer was not quantified in this study.

In the longitudinal scanning, muscle motion in the frontal body plane is indicated. In transverse scanning, muscle motion in the transverse body plane contributes to angle changes. The measured angles and their change arise from the motion of both, the muscle and the US transducer. The study results indicate that the combination of both motions leads to larger angle changes in the transverse compared to the longitudinal scanning plane.

Bulging of the hip abductors occurs in the frontal and the transverse body planes. Bulging appears to be stronger in the transverse body plane as more transducer rocking has been measured in transverse.

This result may be explained by the stronger curvature of the gluteus medius surface in the transverse body plane. The gluteus medius and minimus muscles consist of layers of different depths on the iliac bone. These muscle layers are not parallel with the skin surface and not uniform in thickness. The largest muscle bulk, where the transducer is positioned, is the area where most muscle bulging during activity is expected. Imagining the bulging of gluteus medius in the transverse section of the hip abductors in Figure 4.10 provides an idea of the effect on the transversely orientated transducer.

The strong curvature of the gluteus medius surface and its obliquity relative to the skin surface (Figure 4.10) explains the difficulty to align the hip abductor muscles perpendicular to the US transducer in transverse scanning. A perpendicular arrangement is easier to achieve in longitudinal scanning when the short elevation plane is on the most curved muscle part.

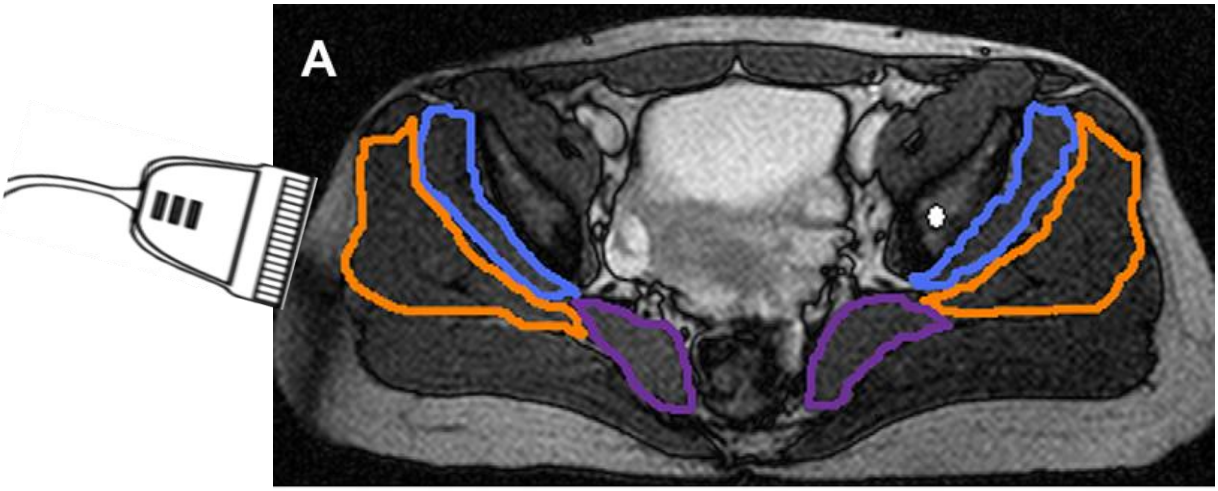


Figure 4.10. MRI section indicating the oblique arrangement of gluteus medius and minimus which is not parallel to the skin surface[3]. In transverse, the transducer is easily tilted by gluteus medius change in shape during activity. Note that in the studies presented here subjects were positioned on a cushioning with an excavation for the buttock to reduce the adipose tissue in the scanning region.

4.2.5 Limitations

Study Two has several limitations. The sample was small with five subjects; however it was sufficient to indicate significant differences in the angles between planes. The study did not include reliability of measurements, because it was assumed that the reliability of the angle measurements would not differ between the scanning planes. The strongest limitation is that transducer motion was not recorded directly and in all three dimensions by a kinematic method (Whittaker et al 2010, Whittaker et al 2009), but was assessed indirectly. Limited to the imaging plane, the M-mode beam gives an indication of the transducer angle that is influenced also by motion of the interfaces relative to which the sound beam angles were measured. Transducer motion out of the scanning plane, tilting of the transducer in the elevation plane, is not assessable by the sound beam. In spite of these limitations, the documented inconsistency in the measurement angles provides a clear argument against the use of the transverse scanning plane for thickness measurements on the hip abductors.

[3] Reprinted from Grimaldi A, Richardson CA, Stanton W, Durbridge G, Donnelly W, Hides J (2009b) The association between degenerative hip joint pathology and size of the gluteus medius, gluteus minimus and piriformis muscles. *Manual Therapy* 14: 605-610. With kind permission from Elsevier (license 2907250984822, 13 May 2012, Appendix 4.1).

4.2.6 Conclusion

In acknowledgement of the limitations, the results indicate an advantage for longitudinal scanning of the hip abductors. The use of thickness measurements of gluteus medius and minimus on longitudinal images is advocated.

4.2.7 Acknowledgement

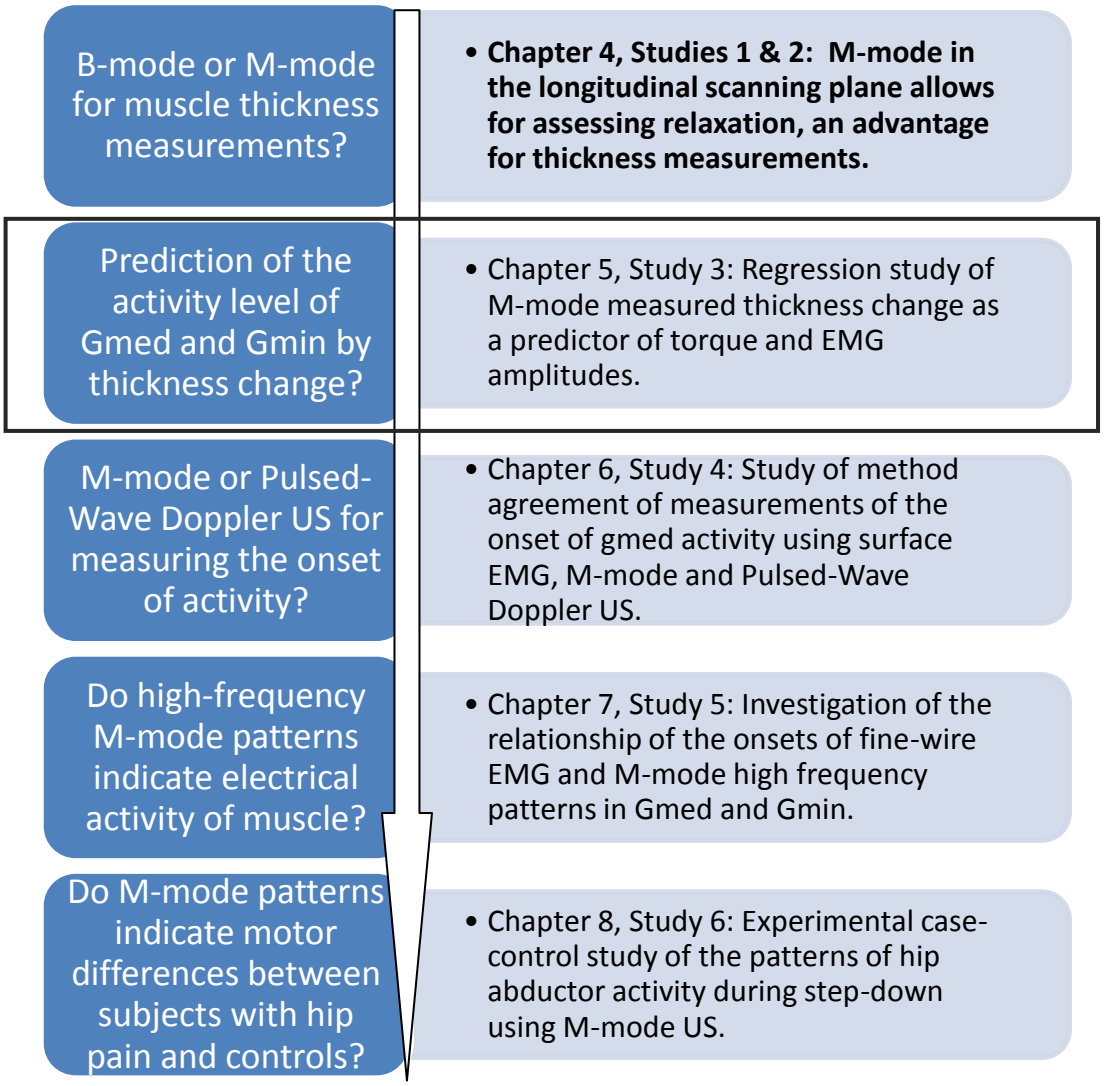
The expert advice on correct terminology of ultrasound transducer planes and motions by Louise Deshon is thankfully acknowledged.

4.3 Summary of the main findings in Chapter Four

Both, B-mode and M-mode US, have unique advantages for the application in muscle thickness measurements. The main advantage of B-mode US is the anatomical orientation and the manual setting of the measurement line perpendicular to the fascias. The main advantage of M-mode US is the ability to indicate motion over time in an image. This feature of M-mode allows for the control of relaxation and also for the investigation of temporal patterns of activity. As muscle activity is dynamic, the M-mode information appears of major advantage.

Measurements of muscle thickness can be performed in the longitudinal and the transverse scanning planes. US scanning of the hip abductors in the longitudinal plane provides the advantages of less transducer motion by muscle bulging and an easier positioning of the transducer perpendicular to the muscles.

In the next research step, M-mode thickness measurements are applied to determine the level of gluteus medius and minimus activity.


B-mode or M-mode for muscle thickness measurements?
• Chapter 4, Studies 1 & 2: M-mode in the longitudinal scanning plane allows for assessing relaxation, an advantage for thickness measurements.
Prediction of the activity level of Gmed and Gmin by thickness change?
• Chapter 5, Study 3: Regression study of M-mode measured thickness change as a predictor of torque and EMG amplitudes.
M-mode or Pulsed-Wave Doppler US for measuring the onset of activity?
• Chapter 6, Study 4: Study of method agreement of measurements of the onset of gmed activity using surface EMG, M-mode and Pulsed-Wave Doppler US.
Do high-frequency M-mode patterns indicate electrical activity of muscle?
• Chapter 7, Study 5: Investigation of the relationship of the onsets of fine-wire EMG and M-mode high frequency patterns in Gmed and Gmin.
Do M-mode patterns indicate motor differences between subjects with hip pain and controls?
• Chapter 8, Study 6: Experimental case-control study of the patterns of hip abductor activity during step-down using M-mode US.

5 Chapter Five: Gluteus Medius but not Minimus Activity Level is Indicated by Thickness Change

Overview of Chapter Five

- Section 5.1 presents Study Three in which the prediction of the level of gluteus medius and minimus activity, as indicated by dynamometry and EMG, from the percentage of change of muscle thickness measured by M-mode US is examined.
- Section 5.2 summarizes the main findings of the Chapters Four and Five for US thickness measurements on muscles and discusses their implications.

5.1 Study Three: Ultrasound measured change of muscle thickness: does it indicate the level of individual hip abductor activity?

5.1.1 Introduction

Differential activity between muscles, muscle parts and also between levels of depth of the hip abductor complex has been documented in asymptomatic subjects (Kumagai et al 1997, Soderberg and Dostal 1978). In early hip joint pathology, a redistribution of hip abductor activity, specifically of force sharing between superficial and deep abductors, has been suggested (Grimaldi et al 2009b, Sims et al 2002), commensurate with current theory on motor adoptions to painful conditions (Hodges and Tucker 2011). Current methods to assess force sharing between superficial and deep muscle rely on invasive fine-wire EMG, which is limited in clinical applicability (Chapter Two, sections 2.3.4. and 2.3.5). In US images, muscle activity is indicated by changes of fascicle length (Chleboun et al 2007, Reeves and Narici 2003), pennation angle (Blazevich et al 2003, Hodges et al 2003, Herbert and Gandevia 1995), cross-sectional area (Delaney et al 2010), muscle shape(Lee et al 2007b, Rankin et al 2005) and muscle thickness (Teyhen et al 2009, Mannion et al 2008b, Kiesel et al 2007b, Hides et al 2006). The applicability of these US measurements differs. Measurements of cross-sectional area and muscle shape are confined to muscles in which the cross-section fits into the field of view of a linear US transducer. Measurements of fascicle length and pennation angle require the determination of the anatomical fascicle plane (Bénard et al 2009) and images of high quality, limiting the measurement to superficial muscles or slender subjects. The measurement with the broadest applicability is muscle thickness.

The relationship between the level of isometric activation and the change of muscle thickness is different between muscles. A curvilinear relationship has been described for the biceps, brachialis, tibialis anterior, transversus abdominis and obliquus internus muscles (Shi et al 2008, Hodges et al 2003). A linear relationship has been described for the lumbar multifidus (Kiesel et al 2008), transversus abdominis (McMeeken et al 2004) and obliquus externus muscles, but for the latter only in isometric trunk rotation (John and Beith 2007). An inconsistent thickening behaviour of obliquus externus has been demonstrated during abdominal hollowing (John and Beith 2007, Hodges et al 2003). Only minimal, curvilinear thickening has been observed in

rectus femoris (Delaney et al 2010). Constant thickness during graded isometric activity has been reported for the gastrocnemius medialis muscle (Narici et al 1996).

Intra-abdominal pressure (Hodges 2005), increasing pressure of adjacent muscles (Delaney et al 2010) and differences in connective tissue elasticity (Hodges 2005) have been discussed as causing diverse thickening behaviours in muscles. A consequence of the differential thickening behaviour of muscles is that the relationship between activity level and thickness change needs to be determined for a muscle before thickness change can be interpreted.

This investigation aimed to determine the predictability of the level of gluteus medius and minimus activity by the change of muscle thickness.

5.1.2 Materials and Methods

The main data set was collected in Setup One using surface EMG (Chapter Three, section 3.3). Setup One provided the larger sample, allowed for comparing the level of torque to gluteus medius and minimus thickness change and the amplitude of surface EMG to gluteus medius thickness change. Additional data for comparing EMG amplitude to gluteus minimus thickness change were provided by the smaller, asymptomatic subsample of Setup Two using fine-wire EMG (Chapter Three, section 3.4). The detailed methods of data collection of both setups are described in Chapter Three. The following paragraphs describe briefly the methods of Setup One and state the differences between the two setups.

5.1.2.1 Subjects

Volunteers were recruited in the School of Physiotherapy, Curtin University. Exclusion criteria were hip pain, history of hip pathology or general musculoskeletal disease. Ethical approval was obtained from the Human Research Ethics Committee, Curtin University (PT0107/2007 and HR143-2009). Subjects provided informed consent.

5.1.2.2 Task and Procedure

In supine lying, isometric abduction of the right hip was performed with contact of the distal lateral thigh against a dynamometer that was fixed to the plinth (Figure 3.9). A randomized sequence of abduction trials in four activity levels was performed in a fully standardized procedure, which included three repetitions of 20%, 40% and 60%

of maximal voluntary isometric contraction (MVIC) and two repetitions of 80% MVIC, sustained for 4 s, with relaxation times of 10 s, 20 s, 30 s and 45 s, respectively. A custom-programmed application (LabVIEW V8.2.1, National Instruments, Texas) informed about trial sequence and provided visual feedback of the actual and the target force levels to the subjects.

5.1.2.3 Dynamometry

Isometric abduction torque was measured by a dynamometer (Mecmesin AFG-500N, Slinfold, UK) with 1000 Hz sampling frequency. Signals were low-pass filtered with a zero lag 4^{th} order Butterworth filter at 40 Hz. The root mean square (RMS) torque amplitude of the force signals was determined in a 0.5 s window during sustained activity. The RMS window was determined choosing a stable phase during sustained activity. Subjects tended to overshoot the low force level and to undershoot the high level. Therefore, RMS torque amplitude was normalized to a medium level activity, the mean of the three 40% trials.

5.1.2.4 Ultrasound imaging

Hip abductor activity was recorded using M-mode US on an Antares 4.0 system (Siemens Medical Solutions, USA) with a 4-9 Hz linear probe (VFX9-4), 38 mm footprint. A custom-programmed US application was set with high contrasts to achieve good distinction of the hyperechoic fascicle and tendon interfaces against the hypoechoic contractile tissue. The M-mode beam was set slightly cranial of the hip joint capsule (Figure 5.1 a). US recordings were captured at 25 frames per second onto digital video (Panasonic digital video camera, NV-MX 500A, Secaucus, USA).

5.1.2.4.1 Scanning location and transducer fixation

The middle to anterior gluteus medius and minimus of the right hip were scanned in the longitudinal plane. The scanning location was determined on the lower half of a line that connected the tip of the greater trochanter to the anterior quarter of a line between the anterior and posterior iliac spines (Figure 5.1). The transducer was housed in a foam block excavated to provide 20° dorsal tilt.

5.1.2.4.2 Ultrasound measurements

Using ImageJ software (version 1.40; rsb.info.nih.gov/ij/), muscle thickness was measured from the inner edges of the fascias (Whittaker 2007, p. 99) off-line during relaxation and the stable phase of sustained activity. The assessor was blinded as to the

level of muscle activity. For each subject, randomly ordered 'stacks' of images of relaxation and of images of sustained activity were enlarged to 200%. In measurements of thickness change, the measurement errors of the relaxed and the activated thickness measurements accumulate. To reduce error summation, averaged relaxed thickness was used as baseline reference for thickness change.

5.1.2.5 Electromyography

EMG was recorded on an Octopus AMT-8 EMG system (Bortec Electronics Inc., Calgary, Canada) at a sampling rate of 1000 Hz using pre-gelled, Ag-AgCl surface electrodes (3M). The ground electrode was attached to the lateral ribs. Following skin preparation, electrodes were positioned with 22 mm inter-electrode distance proximal of the space for the US transducer on a reference line (section 5.1.2.4.1 and Figure 5.1). The raw signals were amplified, 16 bit digitized and displayed on-screen (LabVIEW V8.2.1, National Instruments, Texas). The RMS EMG amplitude of the demeaned and zero-lag band-pass filtered (10 - 400 Hz) signals was determined in a 0.5 s window during sustained activity, allowing for up to 100 ms electromechanical delay relative to the time-window of RMS torque amplitude. RMS EMG amplitude was also normalized to the mean of the 40% trials.

5.1.2.6 Synchronization

Synchronization of torque, EMG and US data was achieved by splitting the signal of a manual trigger to start synchronously the dynamometer and EMG recordings and to create a visual signal on the concurrent M-mode trace (Event Synchronization Unit, PEAK Performance Technologies Inc., Centennial, CO, USA).

5.1.2.7 Method differences in Setup Two including fine-wire EMG

For safety reasons, fine-wire-electrode insertion was restricted to the 'von Hochstetter triangle', an area void of larger nerves and vessels and slightly cranial and ventral to the recording area of Setup One (Figures 5.1 and 3.10). At this location, the hip joint capsule is not scanned, nor another anatomical landmark. The M-mode beam was positioned in the right half of the image corresponding to the main muscle bulk (Figure 5.1 b). Fine-wire EMG signals were band pass-filtered between 10 – 900 Hz. A Toshiba Xario XG system (Toshiba, Medical Division, Australia) with a 7.5 MHz linear probe was used for US imaging. A custom-made triangular support (Figures 3.13 and

3.14) served for transducer fixation. Frames of relaxation and sustained activity were sampled on the US system and stored in tiff format for processing.

It was noted that with fine-wire electrodes inserted, subjects were reluctant to exert full force in the MVIC trials. Therefore, the 80% force level in Setup One was compared to the maximal activity level in Setup Two. Activity levels of 20%, 40% and 60% of individual maximum were performed from low to high.

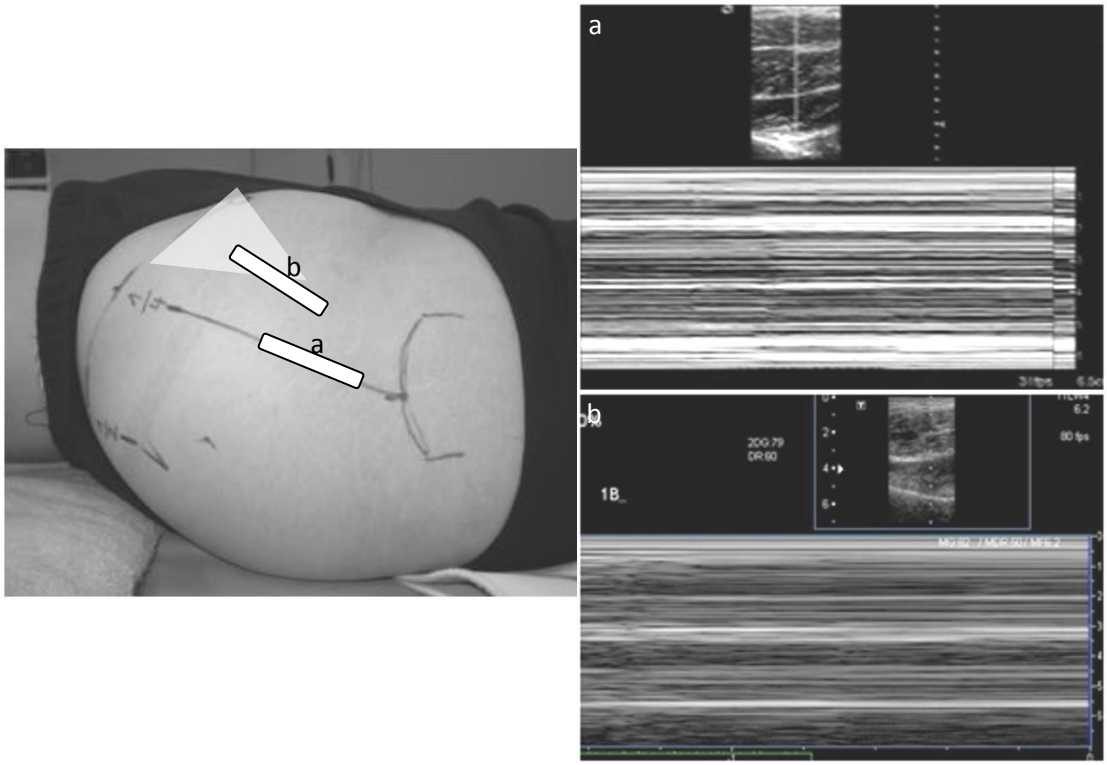


Figure 5.1. US probe locations and M-mode traces of sustained activity: a, Setup One; b, Setup Two; triangular shape: von Hochstetter triangle.

5.1.2.8 Statistics

Intra-tester reliability of measuring the same scans of relaxed and activated muscle thickness was established on randomly chosen images, 87 (53%) of Setup One and 42 (76%) of Setup Two, by $ICC_{3,1}$, mean difference, standard deviation (SD), standard error of measurement (SEM) and minimal detectable change (MDC) (Donoghue et al 2009). There was a two week interval between re-reading the same images. Intra-subject variation of measurements of relaxed muscle thickness was assessed by SD, SEM and MDC. Absolute thickness change in high level activity, which was 80% MVIC in Setup One and maximal activity in Setup Two, was described in mm,

as the percentage of relaxed thickness and relative to intra-subject MDC. Pearson correlation coefficients *r* of torque and EMG amplitudes with percent change of gluteus medius and minimus thickness were determined. Prediction of torque and EMG amplitude from percent change of gluteus medius and minimus thickness was estimated using the determination coefficient r^2 and the standard error of the estimate (SEE), comparing linear, exponential and logarithmic regression models. Significance was set to $\alpha = 0.05$. Statistics were calculated using SPSS (PSAW Statistics v.18.0.0, IBM Corporation, Armonk, New York, USA). Results are presented as mean (SD) if not stated otherwise.

5.1.3 Results

The results are presented separately for gluteus medius and minimus in the text, but they are assembled in the tables to allow for an easier comparison.

5.1.3.1 Data inclusion and reliability

Fifteen subjects (nine females) aged 28 (8) years, with 173 (7) cm height and 67.1 (10.8) kg weight were included in the analysis from Setup One. The data of the sixteenth subject could not be included because the M-mode trace of full activity did not include full muscle thickness. Six subjects (one female) aged 39 (8) years, with 179 (9) cm height and 75.0 (12.4) kg weight were included from Setup Two. The remaining subjects of Setup Two could not be included due to hip pain (four subjects) or unclear demarcation between gluteus medius and minimus in the anterior portion (Bianchi and Martinoli 2007, p. 570) (four subjects). Table 5.1 indicates data inclusion between 90% and 100%. Table 5.2 demonstrates high reliability of thickness measurements with ICCs between 0.857 and 0.962.

Using M-mode US, muscle motion during the relaxation phase was detected in 19 trials (12%) for gluteus medius and, notably, in 35 trials (21%) for gluteus minimus. In these trials, the baseline value was replaced by the mean relaxed thickness value.

Table 5.1. Data inclusion, included data in bold.

	Setup One 165 trials, n=15		Setup Two 66 trials, n=6	
Comparison of muscle thickness change and torque amplitude	Gmed thickness (active) not measurable in 2 trials, =**163 trials (98.8%)**	Gmin thickness (active) not measurable in 1 subject + 5 trials, =**149 trials (90.3%)**	Ok, =**66 trials (100%)**	Gmin thickness (active) not measurable in 1 trial, =**65 trials (98.5%)**
Comparison of muscle thickness change and EMG amplitude	Gmed EMG artefactual in 1 subject + 2 trials, =**150 trials (90.9%)**	n.a.	Gmed EMG artefactual in 2 trials, =**63 trials (95.5%)**	Gmin EMG artefactual in 1 trial, =**65 trials (98.5%)**

Abbreviations in both tables: Gmed, gluteus medius; Gmin, gluteus minimus

Table 5.2. Intra-tester reliability of single M-mode thickness measurements of the relaxed and activated gluteus medius and gluteus minimus muscles, images obtained during same session.

	Gmed Setup One, 87 trials	Gmed Setup Two, 42 trials	Gmin Setup One, 87 trials	Gmin Setup Two, 42 trials
Relaxed: $ICC_{3,1}$, (C.I.)	0.957, (0.935 – 0.986)	0.927, (0.868 - 0.960)	0.933, (0.896 - 0.957)	0.890, (0.804 - 0.939)
Difference relaxed, mm mean, (SD), SEM, *MDC*,	-0.05 (1.00), 0.71, *1.96*	-0.06 (0.64), 0.46, *1.26*	-0.01 (1.37), 0.97, *2.68*	0.16 (0.61), 0.43, *1.20*
Activated: $ICC_{3,1}$, (C.I.)	0.962, (0.943 - 0.975)	0.950, (0.911 - 0.972)	0.884, (0.828 - 0.923)	0.857, (0.751 - 0.920)
Difference active, mm mean (SD), SEM, *MDC*,	0.01 (1.03), 0.73, *2.02*	0.28 (1.01), 0.72, *1.99*	0.18 (0.81), 0.57, *1.59*	0.26 (0.91), 0.64, *1.78*

5.1.3.2 Change of *gluteus medius* thickness with increasing *torque*

In Setup One, US thickness measurements indicated that gluteus medius thickened in 80% MVIC by 5.0 (2.5) mm, which is 20.6% of relaxed thickness. (Figure 5.2 and Table 5.3) The MDC of intra-subject variation was 1.0 (0.3) mm (Table 5.3). In all fifteen subjects, the percentage of change of gluteus medius thickness was significantly correlated with torque, r = 0.80 (0.11). Preferential regression models were in seven subjects linear, in seven exponential and in one logarithmic. Logarithmic regression was not applicable in six subjects due to negative values. R^2 was 0.66 (0.17) for a linear and 0.65 (0.16) for an exponential regression model (Tables 5.5 and 5.6). The power of linear regression analysis was estimated 0.896 using the method proposed by Faul et al (2009).

In Setup Two, five subjects confirmed these results with a significant positive correlation of gluteus medius thickness and torque (Table 5.4)

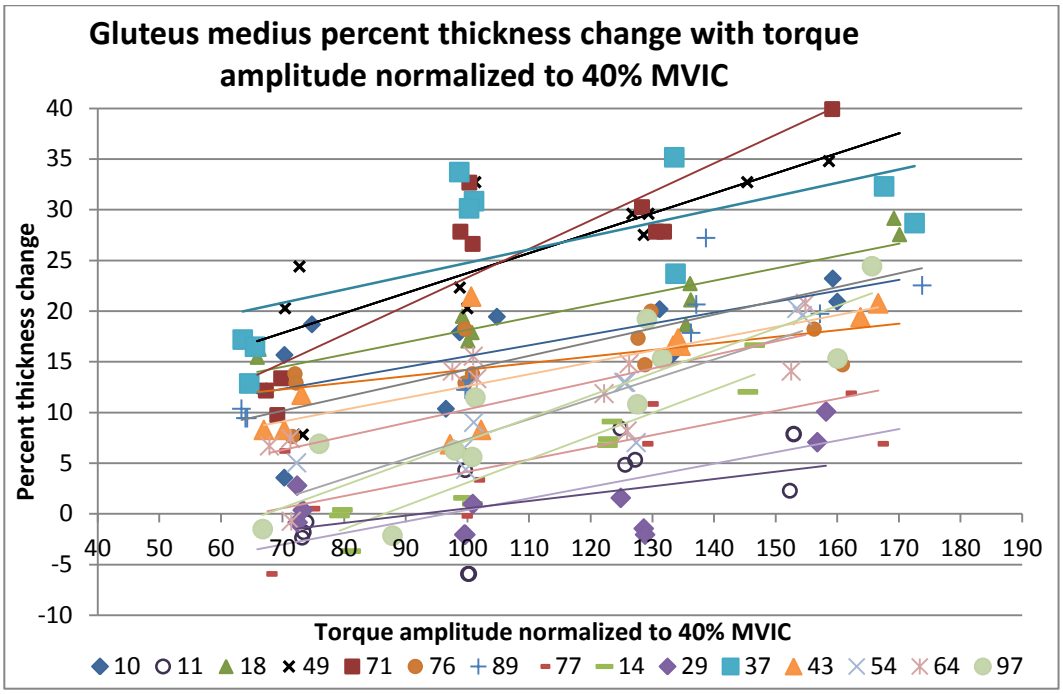


Figure 5.2. XY-plot of the percentage of gluteus medius thickness change with increasing torque, linear regression; data of Setup One.

5.1.3.3 Change of *gluteus minimus thickness* with increasing *torque*

In Setup One, US thickness measurements indicated that gluteus minimus became thinner by 0.2 (1.9) mm, 1.1% of relaxed muscle thickness, in 80% MVIC (Figure 5.3 and Table 5.3). The intra-subject MDC of relaxed gluteus minimus thickness was 0.8 (0.4) mm. Correlation between the percentage of thickness change and torque was weak in thirteen of fourteen subjects, r = 0.05 (0.36). R^2 was 0.12 for linear and 0.11 for exponential regression (Table 5.5). Logarithmic regression was applicable in only one subject.

In Setup Two, gluteus minimus became thinner with maximal activity by 1.6 (1.6) mm, 8.2% (7.9%), (Table 5.4). Intra-subject MDC of relaxed thickness was 0.8 mm. All six subjects indicated a negative correlation, r = -0.66, which was significant in four subjects. R^2 was 0.46 (0.2) for a linear and for an exponential regression model (Table 5.6).

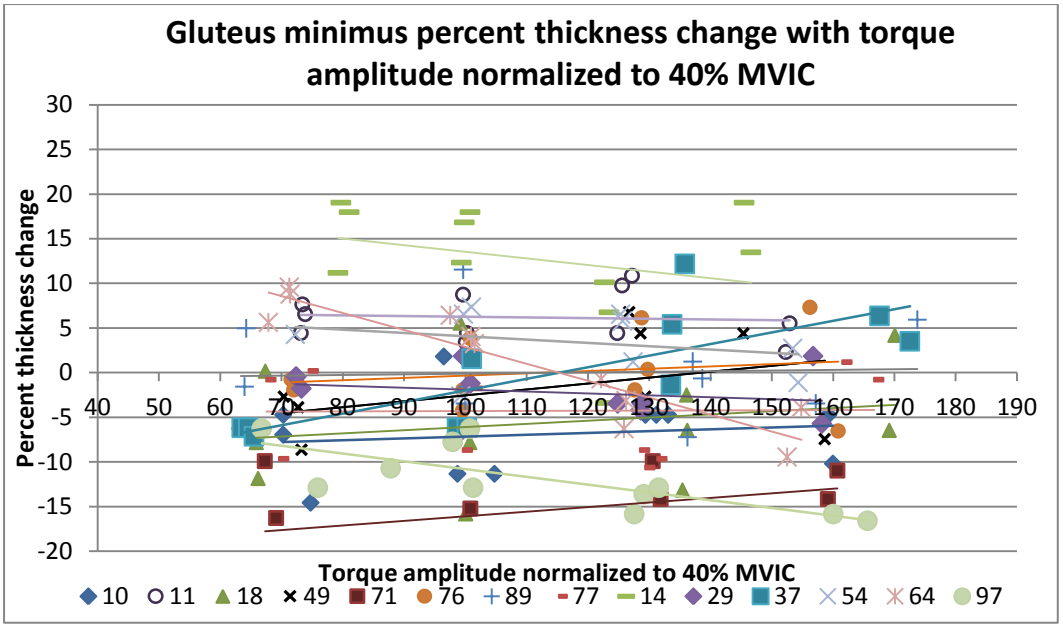


Figure 5.3. XY-plot of the percentage of gluteus minimus thickness change with increasing torque, linear regression; data of Setup One.

Table 5.3. Gluteus medius (Gmed) and minimus (Gmin) thickness change with 80% activity, intra-subject variation of measurements of relaxed muscle thickness; Setup One.

subject	Relaxed Gmed thickness mm	Gmed thickness 80% MVIC mm	Difference mm	Gmed thickness variation in relaxation, SD, SEM, *MDC,* mm	Relaxed Gmin thickness mm	Gmin thickness 80% MVIC mm	Difference mm	Gmin thickness variation in relaxation, SD, SEM, *MDC,* mm
10	22.8	27.8	5.0	0.7, 0.5, *1.4*	15.7	14.6	-1.2	0.2, 0.1, *0.4*
11	32.4	34.1	1.7	0.4, 0.3, *0.7*	15.6	16.2	0.6	0.2, 0.2, *0.4*
14	21.9	25.1	3.1	0.4, 0.3, *0.8*	11.3	13.2	1.8	0.4, 0.3, *0.8*
18	29.2	37.5	8.3	0.7, 0.5, *1.4*	17.6	17.4	-0.2	0.5, 0.4, *1.0*
29	21.5	23.7	2.2	0.7, 0.5, *1.4*	17.3	17.1	-0.2	0.2, 0.1, *0.4*
37	21.3	29.3	8.1	0.8, 0.5, *1.5*	15.8	16.6	0.8	0.3, 0.2, *0.6*
43	22.5	27.0	4.5	0.4, 0.3, *0.7*	24.4	28.6	4.2	0.4, 0.3, *0.9*
49	26.0	34.8	8.8	0.6, .0.4, *1.1*	22.7	22.4	-0.3	0.7, 0.5, *1.3*
54	24.6	29.5	4.8	0.4, 0.3, *0.8*	21.2	21.4	0.2	0.5, 0.4, *1.0*
64	20.3	23.4	3.0	0.5, .0.4, *1.0*	17.3	14.5	-2.7	0.7, 0.5, *1.4*
71	22.3	31.4	9.0	0.4, 0.3, *0.9*	25.4	22.2	-3.2	0.4, 0.3, *0.8*
76	26.7	31.1	4.4	0.5, 0.3, *0.9*	20.4	20.5	0.1	0.2, 0.1, *0.4*
77	26.7	29.2	2.5	0.7, 0.5, *1.4*	19.4	19.4	0.0	0.4, 0.3, *0.8*
89	22.8	27.7	4.8	0.5, 0.6, *1.1*	22.6	23.1	0.5	0.7, 0.5, *1.4*
97	23.1	27.6	4.5	0.4, 0.3, *0.7*	19.8	16.4	-3.4	0.0, 0.0, *0.1*
av	**24.3**	**29.3**	**5.0**	**0.5, 0.4, *1.0***	**19.1**	**18.9**	**-0.2**	**0.4, 0.3, *0.8***

Abbreviations for tables 5.3 – 5.6: av, averaged; SD, standard deviation; SEM, typical error (Hopkins 2000); MDC, minimal detectable change (Donoghue et al 2009); MVIC, maximal voluntary isometric contraction.

Table 5.4. Gluteus medius and gluteus minimus thickness change with maximal activity per subject, intra-subject variation of measurements of relaxed muscle thickness; Setup Two.

subject	Relaxed Gmed thickness mm	Gmed thickness 80% MVIC mm	Difference mm	Gmed thickness variation in relaxation, SD, SEM, *MDC,* mm	Relaxed Gmin thickness mm	Gmin thickness 80% MVIC mm	Difference mm	Gmin thickness variation in relaxation, SD, SEM, *MDC,* mm
Fw3	26.3	29.5	3.3	0.4, 0.3, *0.7*	17.5	17.9	0.4	0.3, 0.2*, 0.5*
FW10	21.8	30.5	8.6	0.3, 0.2, *0.6*	22.4	18.1	-4.3	0.4, 0.3, *0.7*
Fw11	20.7	18.4	-2.4	0.5, 0.4, *1.1*	19.4	18.0	-1.4	1.0, 0.7, *1.9*
Fw15	24.1	32.1	8.1	0.6, 0.4, *1.2*	21.1	19.0	-2.2	0.5, 0.4, *1.0*
Fw5	35.4	36.4	1.0	0.9, 0.7, *1.9*	19.0	18.2	-0.8	0.6, 0.5, *1.3*
Fw8	28.0	35.8	7.9	0.5, 0.3, *0.9*	19.2	17.8	-1.5	0.5, 0.3, *0.9*
av	**26.0**	**30.4**	**4.4**	**0.5, 0.4, *1.1***	**19.8**	**18.2**	**-1.6**	**0.5, 0.4, *1.1***

Table 5.5. Correlation (*r*), significance (*p*), determination coefficients (r^2) and standard error of the estimate (*SEE*) of the percentage of change of gluteus medius (Gmed) and gluteus minimus (Gmin) thickness with torque, Setup One.

subject	Gmed *r*	Gmed *p*	Gmed r^2 linear (*SEE)*	Gmed r^2 exp	Gmin *r*	Gmin *p*	Gmin r^2 linear (*SEE*)	Gmin r^2 exp
10	0.64*	0.033	0.41 (10.7)	0.42	0.15	0.665	0.02 (13.8)	0.03
11	0.65*	0.029	0.43 (9.4)	0.41	-0.08	0.813	0.01 (12.4)	0.00
14	0.94**	0.000	0.88 (3.6)	0.86	-0.27	0.423	0.07 (10.0)	0.09
18	0.92**	0.000	0.85 (6.3)	0.79	0.20	0.557	0.04 (16.0)	0.03
29	0.88**	0.000	0.78 (6.2)	0.73	0.04	0.912	0.00 (13.3)	0.00
37	0.85**	0.001	0.73 (8.7)	0.79	0.85**	0.001	0.72 (8.7)	0.73
43	0.75*	0.008	0.57 (10.0)	0.56	Excluded, only four measurement points			
49	0.77*	0.006	0.59 (8.2)	0.59	0.40	0.292	0.16 (11.9)	0.19
54	0.88*	0.002	0.78 (5.8)	0.71	-0.28	0.463	0.08 (11.9)	0.06
64	0.78*	0.005	0.61 (8.1)	0.63	-0.34	0.308	0.12 (12.2)	0.11
71	0.91**	0.000	0.83 (5.9)	0.86	0.37	0.293	0.14 (13.7)	0.09
76	0.61*	0.046	0.37 (10.8)	0.42	0.18	0.620	0.03 (13.0)	0.04
77	0.70*	0.016	0.49 (10.4)	0.52	0.01	0.968	0.00 (14.6)	0.00
89	0.88**	0.000	0.77 (7.8)	0.76	0.05	0.884	0.00 (16.4)	0.00
97	0.87**	0.001	0.75 (6.9)	0.75	-0.53	0.093	0.28 (11.6)	0.22
av	**0.80**		**0.66 (7.9)**	**0.65**	**0.05**		**0.119 (12.0)**	**0.11**

*significant at 0.05; ** significant at 0.001.
Abbreviations for both tables on this page: av, averaged, exp, exponential.

Table 5.6. Correlation (*r*), significance (*p*), determination coefficients (r^2) and standard error of the estimate (*SEE*) of the percentage of change of gluteus medius and gluteus minimus thickness with torque, Setup Two.

subject	Gmed *r*	Gmed *p*	Gmed r^2 linear (*SEE*)	Gmed r^2 exp	Gmin *r*	Gmin *p*	Gmin r^2 linear (*SEE*)	Gmin r^2 exp
Fw3	0.58*	0.048	0.34 (16.1)	0.44	-0.67*	0.016	0.45 (14.6)	0.46
FW10	0.97**	0.000	0.95 (3.7)	0.87	-0.92**	0.000	0.83 (7.2)	0.81
Fw11	-0.81*	0.003	0.65 (13.0)	0.62	-0.52	0.102	0.27 (18.7)	0.25
Fw15	0.89*	0.002	0.67 (11.0)	0.65	-0.63*	0.051	0.40 (14.9)	0.44
Fw5	0.74*	0.009	0.55 (8.6)	0.48	-0.76*	0.007	0.58(8.4)	0.54
Fw8	0.66*	0.028	0.43 (13.6)	0.51	-0.47	0.143	0.22 (15.8)	0.27
av	**0.49**		**0.60 (11.0)**	**0.60**	**-0.66**		**0.46 (10.9)**	**0.46**

*significant at .05; ** significant at .001

5.1.3.4 Change of *gluteus medius* thickness with *EMG* amplitude

Overall, the relationship of gluteus medius thickness change and EMG amplitude corresponded to the relationship of thickness change and torque, although with slightly smaller r^2.

In Setup One, US measured gluteus medius thickness was positively correlated with surface EMG amplitude, significant in twelve subjects, r = 0.73 (0.20). R^2 was 0.57 (0.24) for a linear and 0.61 (0.24) for an exponential model.

In Setup Two, five of six subjects confirmed the results of Setup One with a significant positive correlation >0.7 of gluteus medius thickness and fine-wire EMG amplitude.

5.1.3.5 Change of *gluteus minimus* thickness with fine-wire *EMG* amplitude

In Setup Two, in five of six subjects gluteus minimus thickness change was negatively correlated with fine-wire EMG amplitude (Figure 5.4), three of them were significant, r = -0.67 (0.22). One subject showed a fair, positive correlation. R^2 was 0.42 (0.28) for a linear and 0.41 (0.28) for an exponential model.

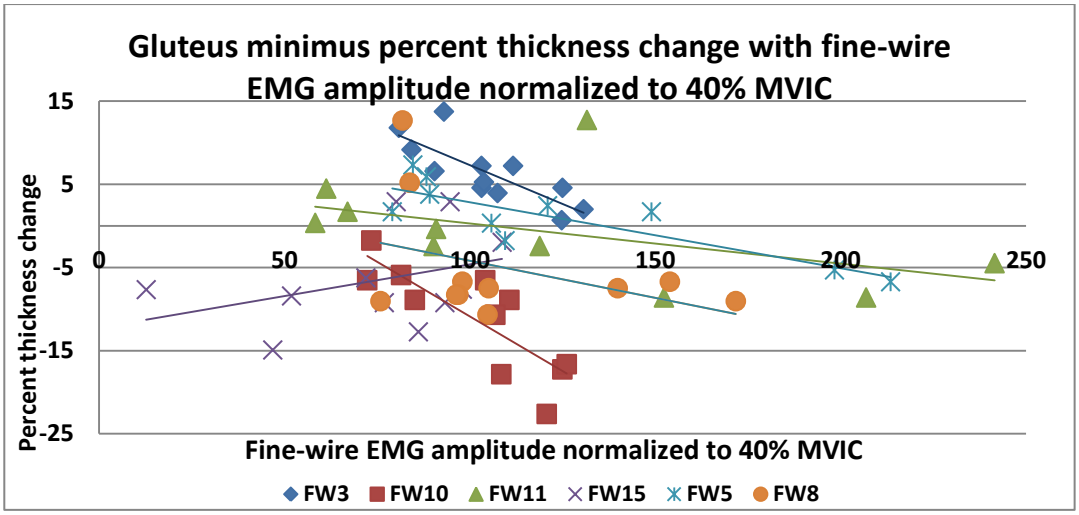


Figure 5.4. XY-plot of the percentage of gluteus minimus thickness change with increasing EMG amplitude, linear regression; Setup Two.

5.1.4 Discussion

This research is the first to investigate the prediction of the activity level of gluteus medius and minimus by the percentage of US measured thickness change. The results document the predictability of gluteus medius activity level by thickness change, however with low precision. Gluteus minimus activity level was not predictable from thickness change. The two hip abductors exhibited a principally different thickening behaviour.

These findings refer to linear regression, which overall yielded the highest prediction. Differences between linear and exponential regression models were below 4%. No model was consistently more effective.

5.1.4.1 Predictability of gluteus medius activity level

Gluteus medius thickening was significantly correlated with torque increase. However, the r^2 and the MDC indicated a low precision of prediction. The determination coefficient r^2 quantifies the proportion of variance of the dependent variable that is explained by the predictor (Portney and Watkins 2000, p. 519). In this study, thickness increase explained 66% of the variance in force and 57% of the variance in EMG

amplitude, with large individual differences. The uncertainty of prediction is in mean 34% for estimation of the exerted torque level. This high uncertainty suggests a low clinical utility of an estimation of the activity level of gluteus medius from thickness measurements. A comparably low prediction of the activity level from thickness change has been reported for the lumbar multifidus, r^2 0.62 (Kiesel et al 2007b), a higher prediction has been found for biceps brachii, obliquus internus, brachialis and transversus abdominis, r^2 0.75 - 0.9 (McMeeken et al 2004, Hodges et al 2003). No prediction was documented for internal and external oblique, r^2 0.02 and 0.05, respectively (Brown and McGill 2010).

A further indicator of low prediction accuracy of gluteus medius activation level is the MDC. The MDC is a statistical estimate of the change that is with 95% certainty larger than measurement error (Donoghue et al 2009), which can be considered 'true' change. The MDC was 20.8% of total gluteus medius thickening in 80% activity, an indication that low levels of activity cannot be distinguished from measurement error. The high MDC indicates that the measurement error of the US thickness measurements was a substantial part of the inaccuracy of prediction. Prediction would be improved when the accuracy of the thickness measurements is improved.

In this study, measurement accuracy is indicated by the intra-tester reliability and by the intra-subject variation of the measurements of relaxed muscle thickness. The intra-tester reliability of the thickness measurements was slightly higher than in the reliability study undertaken in Chapter Four, probably an indication of a better recognition of muscle borders with training by repeated reading. As to literature reports of intra-tester reliability of muscle thickness measurements, the SD and SEM in this study are in the range reported by Mannion et al (2008b) and by Critchley and Coutts (2002), but higher than those reported by Bunce et al (2004) and Hides et al (2007). The SEM of intra-subject variation was lower than the SEM of intra-tester reliability. Likely, an indication for a more consistent recognition of muscle borders in stacked images compared to reading the same image after two weeks interval.

Chapter Four indicated a slightly higher intra-tester reliability of B-mode measured muscle thickness. In spite of a potentially higher measurement precision in B-mode US, the stronger argument may be that using M-mode US, muscle motion during the relaxation phase has been detected in both measured muscles, an indication for invalid measurements of relaxation.

5.1.4.2 Predictability of gluteus minimus activation level

Gluteus minimus thickness change could not be used to predict the activity level. While gluteus medius thickened in mean by 21% with 80% activity, gluteus minimus thickness stayed constant or decreased by 8% during high level activity. The slightly differing results between Setup One and Two may result from the different measurement locations on gluteus minimus. The thickening behaviour of this synergistic pair of hip abductor muscles is different.

5.1.4.3 Differential thickening behaviour in synergistic muscles

Differences in the thickening behaviour of superficial and deep synergistic muscles have been documented for the abdominal muscle group (John and Beith 2007, McMeeken et al 2004, Hodges et al 2003) and the calf muscles. The superficial obliquus externus demonstrated an unpredictable and probably task dependent thickening behaviour (John and Beith 2007) whereas the deeper obliquus internus and transversus abdominis muscles thickened in a linear (McMeeken et al 2004) or curvilinear manner (Hodges et al 2003). The superficial gastrocnemius medialis kept constant thickness while gastrocnemius lateralis and the deep soleus thickened with isometric activity (Managaris et al 1998).

5.1.4.4 Factors which influence the thickening behaviour of the hip abductors

A series of recent studies on the abdominal muscles demonstrated that the mechanical effects of muscle contraction, e.g. muscle thickening, need to be considered within the mechanical environment of a muscle, e.g. the connective tissue attachments (Brown and McGill 2008), the stiffness provided by the extracellular connective tissue matrix (Brown et al 2012) and the morphological and architectural composition within the functional muscle group (Brown et al 2010). Al-Hayani (2009) and Pickard (2005, pp. 45/46) reported gluteus medius attachments to the gluteus minimus fascia, a mechanical connection that may influence the mechanical behaviour. The functional role of these attachments and the possible interaction of gluteus medius and minimus remain to be examined.

Muscle architecture studies allow for some inferences on the influence of gluteus medius and minimus architecture on their thickening behaviour. The change of muscle thickness during activity is explained by the thickening of muscle fibres during contraction (Boyett et al 1991). In parallel muscles, the increase of a fibre's cross-section results in an increase of muscle thickness (McMeeken et al 2004, Bakke et al 1992). In pennate muscles, muscle thickness has been modelled as a function of fascicle length (FL), pennation angle (θ) and the angle γ between lower and upper fascia, assuming straight fascicles in a two-dimensional model:

$$Muscle\ thickness = \frac{Fascicle\ length \times\ sin\ (180° - (\gamma + 180° - \theta))}{sin\ (\gamma + 90°)}$$

(Blazevich et al 2006), which can be simplified to:

$$Muscle\ thickness = Fascicle\ length \times sin\theta$$

in the case of parallel muscle fascias.

If fascicle shortening and the increase in pennation angle compensate for each other, muscle thickness remains constant during activity, as has been documented for the medial gastrocnemius (Narici et al 1996) and as has been assumed in classical models of dynamic muscle architecture (Otten 1988). If the effect of pennation increase exceeds fascicle shortening, the muscle thickens during activity, as has been documented for the soleus muscle (Managaris et al 1998). These literature reports provide evidence that the relationship between fascicle shortening and increase in pennation angle is variable between muscles. In conclusion, if a muscle is thinning, it can be assumed that fibre shortening exceeds the effect of pennation increase.

Gluteus minimus fibres run parallel with the femoral neck and insert into the strong aponeurosis (Gottschalk et al 1989), a unipennate arrangement (Figure 5.5). In isometric activity of gluteus minimus, the direction of fibre pull is oblique and cranial towards the iliac bone, counteracted by the limits of tendon / aponeurosis elasticity and by the elasticity of the connective tissue network. This architectural constellation together with the notion that fibre shortening can counterbalance or probably even exceed thickening by increased pennation may explain constant or even reduced muscle thickness during activity (Figure 5.5).

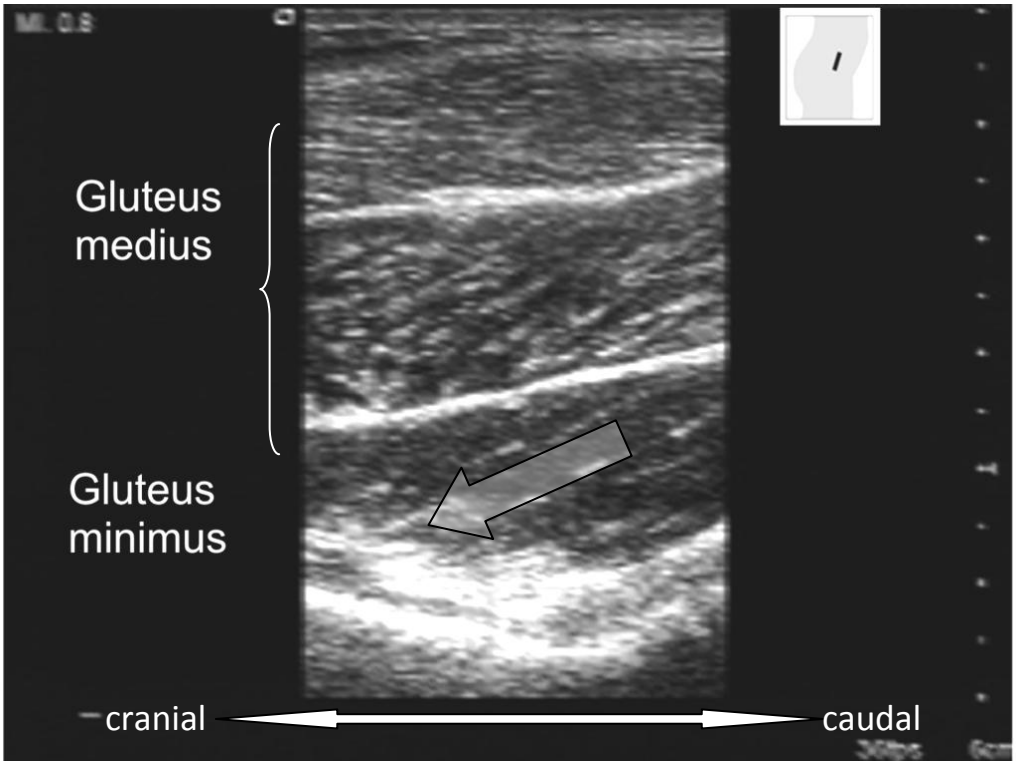


Figure 5.5. Gluteus medius and minimus architecture, direction of pull of gluteus minimus activity (transparent arrow). Note that in the depicted muscle levels the fascicles are delineated only in part, an indication that the image is not taken in the fascicle plane and therefore not suitable for taking valid fascicle measurements (Bénard et al 2009).

5.1.4.5 Limitations

US observed phenomena should be controlled in both scanning planes to avoid a misinterpretation (Zamorani and Valle 2007, p. 47). Even in isometric activity, the muscle moves initially (Karamanidis et al 2005) while the US transducer is stationary. Estimation of the activity level from muscle thickening requires the same part of the muscle being measured during relaxation and activation.

US scanning in the longitudinal plane enabled control of thickness changes by muscle motion within the frontal / coronal body plane. Thickness change by muscle motion in the transverse body plane could be controlled retrospectively for the data of Setup One, in which also transversely scanned data had been collected with a consistent methodology (Chapter Three, section 3.3). The transverse B-mode recordings of isometric hip abduction suggested regular gluteus medius and minimus motion in the transverse plane.

Gluteus medius more than gluteus minimus is a muscle of non-uniform thickness (Grimaldi et al 2009b) (Figure 4.10). Transverse gluteus medius motion brings another muscle section, potentially of different thickness, into the field of view of the US transducer. Transverse gluteus medius motion may have reduced the precision of the prediction of the activity level by thickness change.

In conclusion, in order to measure thickness change during isometric activity in muscles of non-uniform thickness, the operator needs to control initial muscle motion in both scanning planes and determine a measurement location on which the anatomical thickness of the muscle stays constant during the muscle motion that occurs with activity. Comparing both scanning positions, the site which was used in the fine-wire EMG study appears to allow for less transverse muscle motion because it is close to the muscle origins. On the other hand, a lower prediction of torque by thickness change and less gluteus medius thickening in the fine-wire EMG study (Tables 5.3 – 5.6) support that a measuring site close to the bony origins of a muscle offers less space for muscle thickening compared to a site on the main muscle bulk, as chosen in the surface EMG study.

A further study limitation refers to the power of the fine-wire EMG analysis. While the power to detect gluteus medius differences between relaxation and high level activity in Setup One was 0.93, the small sample of Setup Two provided a power of 0.74 to detect differences for gluteus minimus. Therefore, the statement of gluteus minimus thinning in the anterior muscle warrants further examination.

Activity levels were not the same between the two Setups because the willingness of the subjects to perform maximal activity differed with invasive electrodes. This limitation affected the range of measured activity but not the general relationship with thickness change. To enhance the comparability between both Setups, 80% MVIC in Setup One was compared to maximal activity Setup Two.

5.1.5 Conclusions

Study Three indicates important limitations for the application of US thickness measurements to estimate the level of hip abductor activity. The precision of estimating the level of gluteus medius activity by US measured thickness change was low. In comparison to surface EMG, US provided no alternative of clinical interest. Gluteus minimus activity levels were not assessable by thickness change. The current gold

standard for measurements of the level of gluteus minimus activity, fine-wire EMG, cannot be replaced by non-invasive US thickness measurements.

5.1.6 Acknowledgements

I gratefully acknowledge the support of Dr Will Gibson for inserting the fine-wire electrodes. Further, the loan of ultrasound equipment by Siemens Medical Systems, Western Australia and Toshiba, Medical Division, Australia is acknowledged. Technical support by Paul Davey and advice in anatomical questions by John Owens was highly appreciated.

5.2 Ultrasound measurements of muscle thickness change: still a challenge

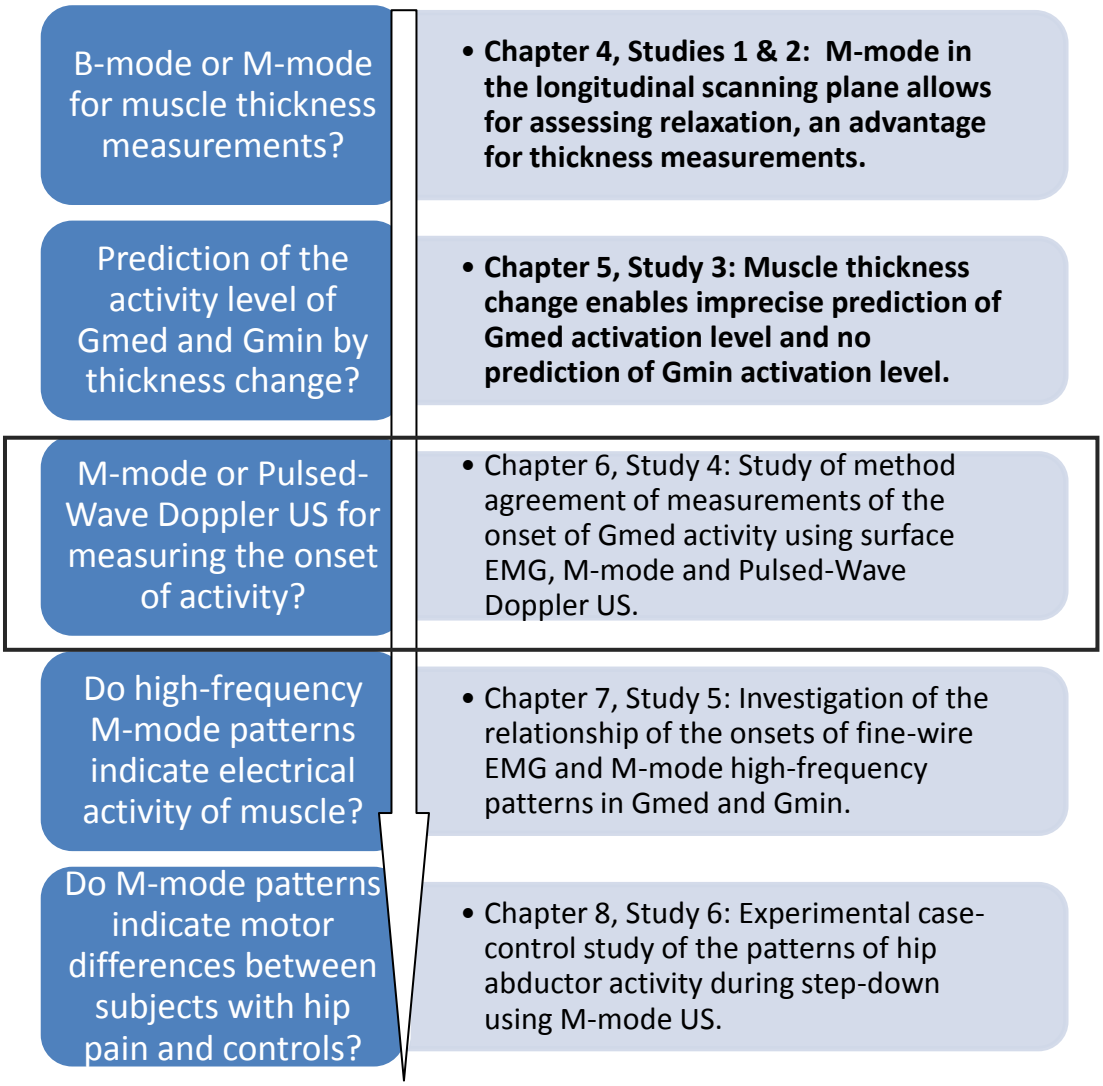
Chapters Four and Five reported the exploration of thickness measurements for estimating the level of gluteus medius and minimus activity. A large part of this presentation was devoted to methodological considerations, a fact, which reflects the uncertainties still linked to the application of US thickness measurements of muscle activity.

The main improvement promised by US imaging in comparison to EMG is the non-invasive assessment of deep muscles. The literature and also this research demonstrate limitations of the assessment of muscle activity by thickness change (Delaney et al 2010, John and Beith 2007, Hodges et al 2003, Narici et al 1996). Not all muscles indicate their activity by measurable thickness change. The relationship between thickness change and activity level is muscle specific and needs to be verified. There is no general rule for interpreting muscle thickness change.

The control and standardization of US image production is a challenge in the application of muscle thickness measurements. The measurement error of US thickness measurements needs to be minimized.
The experience with Studies One to Three suggested the following points:

- a careful setting of dynamic range and contrast in M-mode to facilitate the distinction of fascicles and muscle fascia;
- fascial contours may be enhanced by Edge Enhancement post – processing, an application of high-pass filters to enforce hyperechoic interfaces (Hoskins et al 2010, p. 61);
- the B-mode image aside the M-mode trace to allow for easier anatomic recognition;
- data measurements in a non-compressive image format such as .tif or .bmp (video recording results in a loss of resolution);
- an indication of transducer tilt in the US system, as in a mason's level, would enhance the repeatability of the scanning angle.

The US-provided insight into mechanical changes during muscle activity revealed a lack of knowledge of the influencing factors. The arrangement of muscle fascicles and their dynamic properties may be one influencing factor. The change of muscle shape happens within a complex environment of tissues and organs with different elasticity and resistance. The power of the muscle contraction is guided, splinted and transmitted by the connective tissue network (Huijing 2009, Huijing and Baan 2001). The resulting change of muscle shape may be unexpected and difficult to relate to the underlying quantity of excitation without further knowledge of the mechanical conditions and constraints (Brown and McGill 2010, Delaney et al 2010, Brown and McGill 2008).


B-mode or M-mode for muscle thickness measurements?
• Chapter 4, Studies 1 & 2: M-mode in the longitudinal scanning plane allows for assessing relaxation, an advantage for thickness measurements.
Prediction of the activity level of Gmed and Gmin by thickness change?
• Chapter 5, Study 3: Muscle thickness change enables imprecise prediction of Gmed activation level and no prediction of Gmin activation level.
M-mode or Pulsed-Wave Doppler US for measuring the onset of activity?
• Chapter 6, Study 4: Study of method agreement of measurements of the onset of Gmed activity using surface EMG, M-mode and Pulsed-Wave Doppler US.
Do high-frequency M-mode patterns indicate electrical activity of muscle?
• Chapter 7, Study 5: Investigation of the relationship of the onsets of fine-wire EMG and M-mode high-frequency patterns in Gmed and Gmin.
Do M-mode patterns indicate motor differences between subjects with hip pain and controls?
• Chapter 8, Study 6: Experimental case-control study of the patterns of hip abductor activity during step-down using M-mode US.

6 Chapter Six: M-mode Ultrasound Visualizes the Sequence of Activity-Related Muscle Motion in Different Levels of Depth of the Hip Abductors

Overview of Chapter Six

- Section 6.1 presents Study Four in which measurements of the onset of gluteus medius activity by surface EMG, M-mode and Pulsed Wave Doppler US were compared for measurement reliability and the additional information provided by each US mode.
- Section 6.2 describes an experiment undertaken to compare the temporal accuracy of methods used for synchronizing US with EMG or dynamometry data.
- Section 6.3 discusses the use of the scanning planes in US onset measurements.
- Section 6.4 documents observations of other sources of motion than muscle activity in M-mode US.
- Section 6.5 summarizes the main conclusions of Chapter Six for the application of US onset measurements on the gluteus medius muscle.

6.1 Study Four: Onset of hip abductor activity measured by surface electromyography and ultrasound: M-mode or Pulsed Wave Doppler?

6.1.1 Introduction

Pathology induced alterations in patterns of muscle activity can refer to the force sharing (Grimaldi et al 2009a, Grimaldi et al 2009b) and also to the relative timing of muscle activity (Lucas et al 2010, Hodges and Richardson 1999). The previous two chapters related to the aspect of force sharing. The following chapters refer to the timing of hip abductor activity. The most commonly investigated aspect of timing is the onset of activity.

The gold standard for assessing the onset of muscle activation of single muscles is electromyography (EMG), the limitations of which have already been discussed (Chapter Two, sections 2.3.4 and 2.3.5). US measurements of activity onset are based on the detection of muscle motion (Vasseljen et al 2009, Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006). Valid US measurements of the onset of abdominal, spinal and quadriceps muscle activity have been reported using M-mode, Tissue Doppler or Strain rate US. However, the precise relationship between the onset of electrical excitation and muscle motion depended on the sequence of activation within the muscle group (Vasseljen et al 2009, Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006). A high agreement of electrical and motion onsets was indicated for the first activating muscle, but a more variable relationship for later activating muscles.

Therefore, the interpretation of the onset of muscle motion is dependent on the timing of a muscle's activation relative to adjacent muscles. The onset of muscle motion indicates the onset of electrical excitation when adjacent muscles activate synchronously. Interpretation of muscle motion is less clear in muscles that activate later than an adjacent muscle. In the abdominal and spinal muscle groups, it is known that activity onset differs between superficial and deep muscles (Moseley et al 2002, Hodges and Richardson 1999). The temporal pattern of activity of the superficial gluteus medius and the deep gluteus minimus muscles is not known.

6.1.1.1 Ultrasound imaging modes for measuring muscle motion

The temporal resolution of US B-mode in commonly available US systems is limited to 20 – 40 ms. Observed onset differences, e.g. between abdominal muscles, were > 30 ms, standard deviations of single muscle onsets between 11 and 110 ms (Hodges and Richardson 1999). Higher temporal resolution than in B-mode US, < 5 ms, can be achieved using M-mode or Doppler US modes. The following section provides a short description of the US modes used in Study Four. A more detailed description is provided in Chapter Two, section 2.5.6.

In M-mode US, a single, stationary sound beam interrogates the scanned tissue section and detects the actual depth of sound-reflecting interfaces. Straight, horizontal lines indicate stationary interfaces. When the interface moves, its replicating line changes greyscale, position or is interrupted (Anderson 2007, Vasseljen et al 2006). In the M-mode trace, time is on the *x*-axis and interface depth on the y-axis (Anderson 2007, Hedrick et al 2005, Gent 1997) (Figure 6.1).

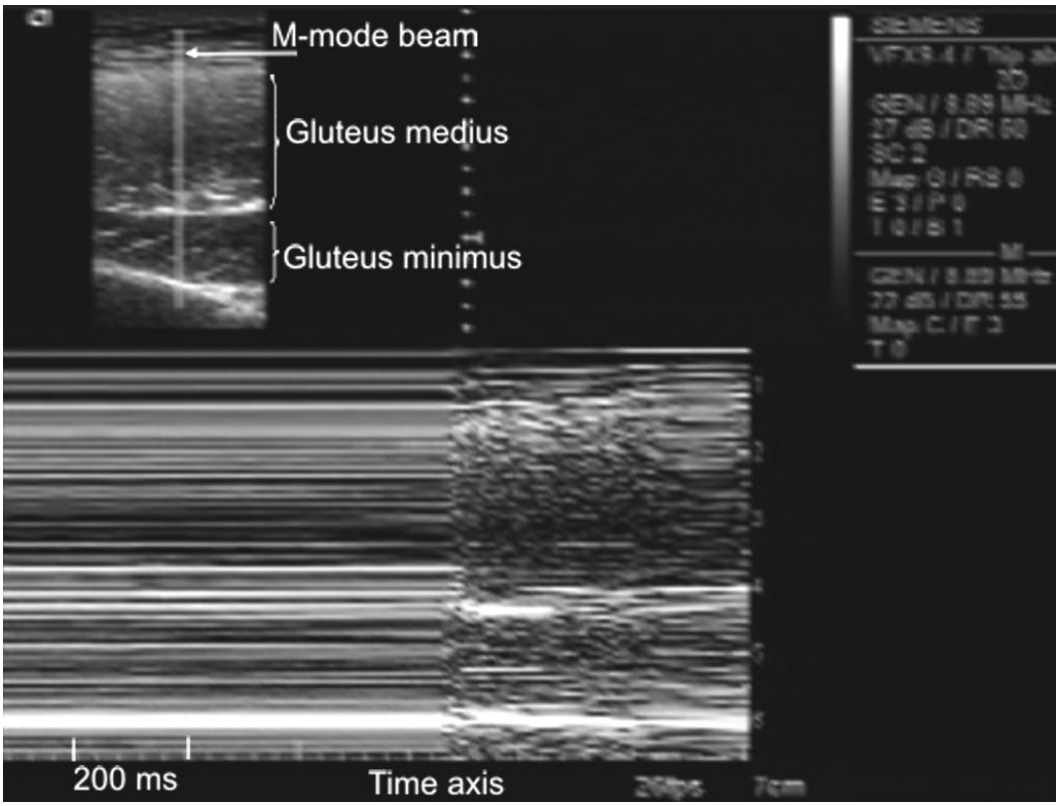


Figure 6.1. M-mode trace of gluteus medius and minimus activity in 60% MVIC.

Also based on a single sound-beam, Pulsed Wave Doppler US detects motion by a frequency analysis of the echoes. Motion of the reflecting interface causes a frequency shift in the echoes, the 'Doppler' effect (Hoskins et al 2010, p. 84-86). The velocity of motion can be calculated from the frequency shift when the angle between the sound beam and the direction of motion is determined. The sensitivity of Doppler US to detect motion is angle dependent and optimal for motion along the US beam (Hoskins et al 2010, pp. 114-116, Anderson 2007, p. 94). In Doppler velocity calculations, an angle of 0° (= motion is aligned with the sound beam) is assumed when the sound beam has not been changed manually. In Pulsed Wave Doppler, only echoes from a small area, the 'gate', are analysed. The Pulsed Wave Doppler image is a spectral display with time on the *x*-axis and the velocity of the measured motion on the *y*-axis (Gent 1997) (Figure 6.2).

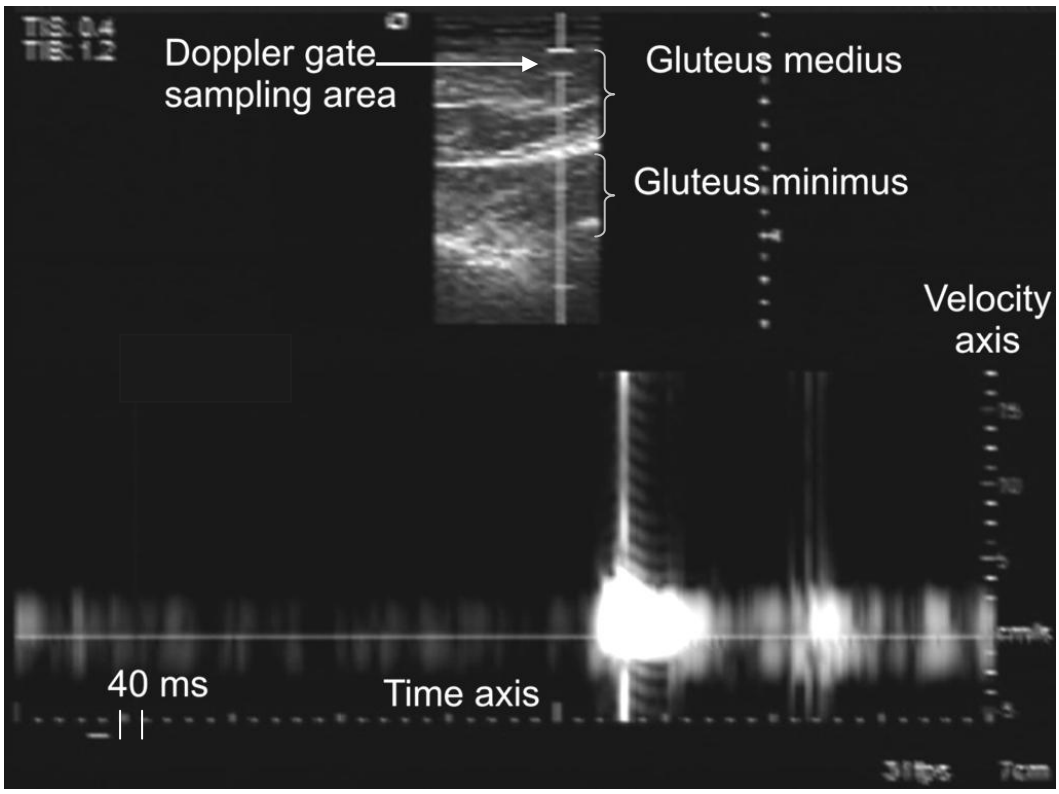


Figure 6.2. Pulsed Wave Doppler display of activity of the superficial gluteus medius in 60% MVIC.

The objectives of Study Four were

(1) to estimate the intra-tester reliability of surface EMG, M-mode and Pulsed Wave Doppler measurements of the onset of gluteus medius activity;

(2) to explore the synchronicity of motion onset in the gluteus medius and minimus muscles; and
(3) to describe the relationship of surface EMG and gluteus medius motion onsets in different force levels.

6.1.2 Methods

In Study Four, data of Setup One were analysed. The detailed methods of data collection are presented in Chapter Three, section 3.3.

6.1.2.1 Subjects

Volunteers were recruited at the School of Physiotherapy, Curtin University. Exclusion criteria were hip pain, history of hip pathology and general musculoskeletal disease. Ethics approval was obtained from the Human Research Ethics Committee, Curtin University (PT0107/2007). Signed, informed consent was obtained prior to data collection.

6.1.2.2 Task and Procedure

Participants performed right isometric hip abduction in 'natural' velocity against a dynamometer. Abduction trials were performed in a fully standardized, screen-guided sequence, in which the target force level, feedback of the exerted force and relaxation phases were visualized. Four series of three repetitions of 20%, 40% and 60% MVIC, and two repetitions of 80% MVIC, sustained 4 s, with relaxation breaks of 10 s, 20 s, 30 s or 45 s, respectively, were performed in random sequence. One series in M-mode and one series in Pulsed Wave Doppler were analysed for this study.

6.1.2.3 Dynamometry

Isometric abduction force was measured as the rise of pressure of the right thigh against the dynamometer (Mecmesin AFG-500N, Slinfold, UK). Signals were collected at 1000 Hz and low-pass filtered with a zero lag 4^{th} order Butterworth filter at 40 Hz.

6.1.2.4 Electromyography (EMG)

Surface EMG of the right gluteus medius was recorded at a sampling rate of 1000 Hz on an Octopus AMT-8 EMG system (Bortec Electronics Inc., Calgary, Canada) with an inter-electrode distance of 22 mm. The electrode position was marked on the

line between the greater trochanter and a point 25% of the distance between the anterior and posterior superior iliac spines (Figures 3.1. and 3.2). This electrode position was more proximal than recommended (Rainoldi et al 2004, Freriks et al 1999) but on the same muscle part as the US[4]. Onset differences has been reported for the gluteus medius parts (Soderberg and Dostal 1978). Raw EMG signals were demeaned, zero-lag band-pass filtered between 10 and 400 Hz with a 2nd order Butterworth filter and full-wave rectified for visual onset detection.

6.1.2.5 Ultrasound

US data were recorded on an Antares 4.0 system (Siemens Medical Solutions, USA) at 9 MHz using a linear probe (VFX9-4), 38 mm footprint. The US transducer was positioned in the longitudinal scanning plane on the lower half of the reference line described above, housed in a foam block that was excavated at a 20° angle and fixed with belts around the pelvis (Mannion et al 2008a, Bunce et al 2002). The transducer angle, gain, TGC and dynamic range were adjusted to achieve a B-mode image that allowed for clear recognition of the iliac periosteum and the gluteus minimus aponeurosis, if achievable, single muscle fascicles. The M-mode cursor was set cranial to the reflected head of rectus femoris at the thickest muscle bulk. Strong contrasts and highest sweep speed were set. The sampling rate was, depending on the image depth, approximately 263 Hz, providing a temporal resolution of 3.8 ms.

Doppler US data were recorded with a pulse repetition frequency of 1221 Hz, a gate width of 5 mm, and the lowest wall filter setting (37 Hz). The Doppler gate was ensured to stay within gluteus medius during activity. Gain was regulated to display no or minimal signal in relaxation. US data of the complete trial series were recorded at 25 frames per second onto video (Panasonic digital video camera, NV-MX 500A, Secaucus, USA).

6.1.2.6 Synchronization

Synchronization between dynamometry, EMG and US data was achieved by a manual trigger that started dynamometer and EMG recording and created a time stamp on the concurrent video frame (Event Synchronization Unit, PEAK Performance Technologies Inc., Centennial, CO, USA). Each video frame included new data from a 40

[4] As described in Chapter Three, section 3.1.1, two pairs of electrodes has been used in Setup One, one pair cranial to the US transducer and a second pair posterior to the US transducer. To reduce the complexity of the presentation, only the electrode configuration with higher concordance to the US data is reported here.

ms period. The time stamp increased in brightness over time. In order to increase the temporal resolution of the synchronization event, it was differentiated between dim and bright time stamps. Of all M-mode trials, 52.9% showed a dim time stamp that was counted for 10 ms, and 47.1% showed a bright time stamp that was counted for 30 ms. The reliability of assigning 'dim' and 'bright' was very good with 99 of 100 trials assigned consistently after one week.

6.1.2.7 Operational definitions of onset

Human performance may include incomplete relaxation, hesitation or pre-emptive activity before activity onset. These behaviours complicate a computed detection of valid onsets (Allison 2003, Hodges and Bang 1996). All onsets were determined visually on-screen, blinded for other onset measurements. Screen resolution was 0.6 mm = 10 ms for EMG and 0.7 mm = 10 ms for US. EMG onset was defined as the instant when a constant signal pattern of minimal or low amplitude changed into a pattern of increasing amplitude. M-mode onset was defined as the first change in greyscale in at least three M-mode lines. Onset in Pulsed Wave Doppler was defined as the beginning of continuing spectral increase over the baseline level.

6.1.2.8 Statistics

Intra-tester reliability of EMG and US onsets of gluteus medius activity was assessed on 40 randomly chosen data sets on which measurements were taken twice with a one week interval. Reliability estimation was by intraclass correlation coefficient ICC $_{3,1}$ and by mean, standard deviation (SD), standard error of measurement (SEM) and minimal detectable change (MDC) (Donoghue et al 2009) of the difference between measuring occasions.

Onset of gluteus medius relative to gluteus minimus motion was described from the M-mode US data by mean, SD and confidence interval (CI).

EMG and US onset are separated by a physiological time lag, the electromechanical delay. This latency was calculated for single and for averaged trials per subject and force level. As skewed onset distributions (Mannion et al 2008a, Pulkovski et al 2008) may result in inappropriately applied statistics, the regular and an alternative statistic method for skewed data are presented. The correlation between EMG and US onsets was estimated by Pearson correlation *r* on the raw and the log-transformed data (Rascati et al 2001, Bland and Altman 1996). The agreement of EMG

and US onsets was estimated by the limits of agreement (LOA). The latencies between EMG and US onset were described by mean and SD, and also by median and interquartile range (IQR).

The influence of the force level was assessed by ANOVA. In order to also determine the influence of the velocity of activation on the latency between EMG and motion onset, activation velocity was determined from the initial 100 ms of dynamometry data. The relationship between activation velocity and onset latency was examined by Pearson correlation *r.* Statistical significance was set to $p < 0.05$.

6.1.3 Results

6.1.3.1 Data inclusion and onset reliability

Subjects were 13[5] healthy volunteers (7 female), aged 29 (SD 7.6) years with a BMI of 22.6 (SD 3.1) kg/m².

Altogether 286 isometric abduction trials were recorded, 143 in M-mode and 143 in Pulsed Wave Doppler. Surface EMG data of 222 trials (77.6%) could be included. EMG data exclusion was due to ambiguous onsets (29 trials, 45% of exclusions), activity was not detected (27 trials, 37% of exclusions) or technical problems with synchronization or baseline data length (10 trials, 16% of exclusions). Notably, 31 (48%) of the excluded trials were in the 20% torque level. US onsets were determined only in trials with included EMG data. In the US data, 8 trials were excluded due to a missing synchronization signal and 1 trial because the baseline was not relaxed. For 213 trials (74.5%) datasets were complete (Table 6.1). Intra-tester reliability is given in Table 6.2.

Table 6.1. Included trials in number and percentage of recorded trials, per force level.

	20% MVIC	40% MVIC	60% MVIC	80% MVIC
M-mode (108 trials)	20 (51.3%)	31 (79.5%)	34 (87.2%)	23 (88.5%)
$P_{ulsed}W_{ave}$ Doppler (105 trials)	18 (46.2%)	32 (82.1%)	31 (79.5%)	24 (92.3%)

[5] In the remaining three subjects of Setup One, the signal to noise ratio was not sufficient for onset detection. Reduced EMG data quality may have been caused by the unusual electrode location close to the muscle origin.

Table 6.2. Intra-tester reliability of onset measurements on the same data repeated after one week.

	$ICC_{3,1}$ (C.I.)	Mean, ms	SD, ms	SEM, ms	*MDC*, ms
EMG	1.000 (1.0 – 1.0)	-0.2	14.9	10.5	*29.2*
M-mode Gmed	1.000 (1.0 – 1.0)	7.6	19.2	13.6	*37.6*
M-mode Gmin	1.000 (1.0 – 1.0)	1.3	10.4	7.4	*20.5*
$P_{ulsed}W_{ave}$ Doppler	1.000 (1.0 – 1.0)	8.6	33.2	23.5	*66.1*
(disregading 1 outlier)		(5.0)	(23.6)	(16.7)	*(46.3)*
Force	1.000 (1.0 – 1.0)	-0.1	6.4	4.5	*12.5*

EMG, electromyography; Gmed, gluteus medius; Gmin, gluteus minimus; ICC, intraclass correlation coefficient; MDC, minimal detectable change.

6.1.3.2 Gluteus medius relative to gluteus minimus motion onset

Gluteus medius motion onset occurred in mean 36 (SD 85, C.I. ± 16) ms later than gluteus minimus motion onset, $p < 0.0001$. Regularly, the M-mode traces indicated a sequential motion onset also between a deep and a superficial muscle level of gluteus medius (Figures 2.7, 6.7, 6.12, 7.1, 7.14).

6.1.3.3 Agreement of EMG and US onsets of gluteus medius activity

Correlation of EMG and M-mode US onsets of gluteus medius activity was r = 0.998 for EMG and Pulsed Wave Doppler onsets. Log transformation of onsets reduced the range and r to 0.984 and 0.0.978 for M-mode and Pulsed Wave Doppler, respectively (Figure 6.3 exemplary for M-mode).

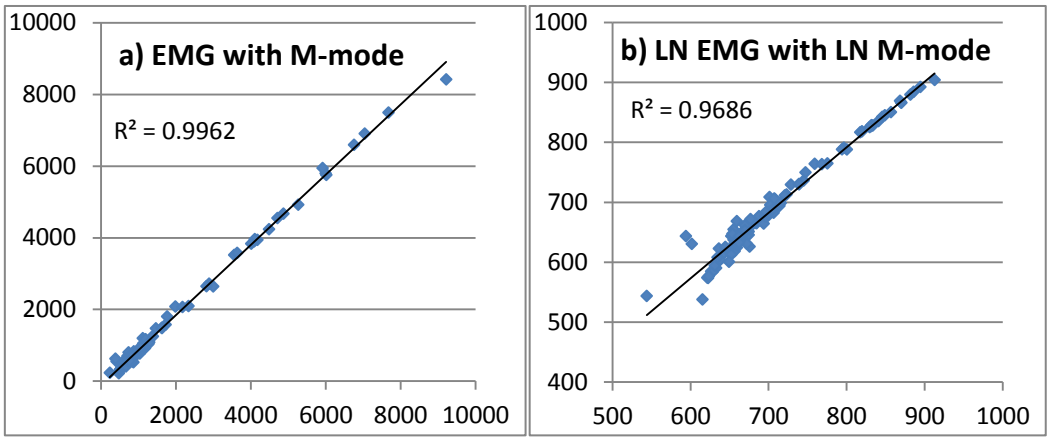


Figure 6.3. XY-plots of gluteus medius activity onsets by EMG and M-mode US, a) raw onsets; b) log-transformed onsets. Onsets were log-transformed to reduce the influence of the large range on correlation.

In spite of the excellent correlation, latency between EMG and US onset varied considerably, leading to wide limits of agreement (Figure 6.4). In 7.5% of M-mode and 9.5% of Pulsed Wave Doppler trials, motion onset was registered before EMG onset.

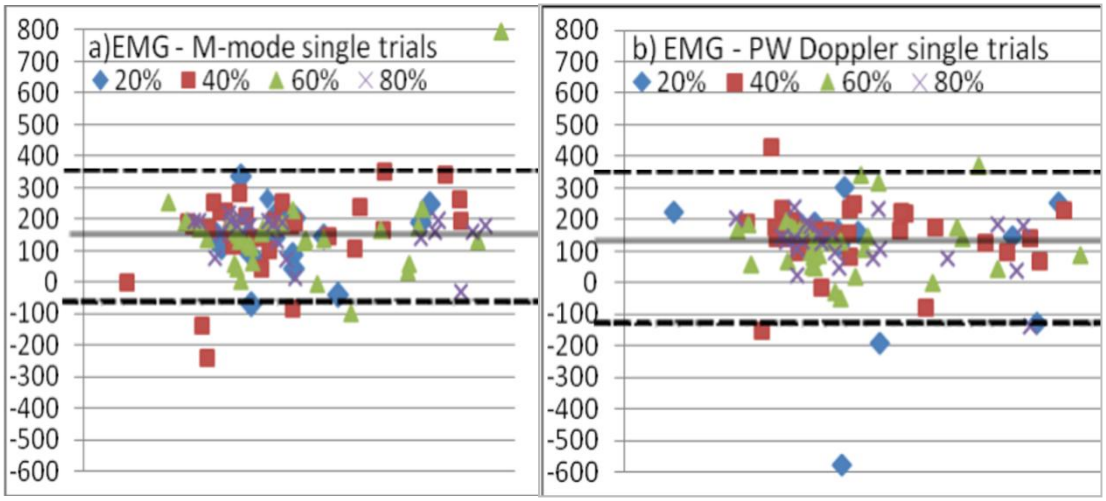


Figure 6.4. Bland-Altman plots of EMG and US onsets. Dashed lines = limits of agreement (LOA) = 1.96 standard deviations, continuous line = mean difference; logarithmic scale of the x-axis to show full range.
a) EMG versus M-mode, mean 152 ms LOA ± 223 ms,
b) EMG versus Pulsed Wave (PW) Doppler, mean 126 ms LOA ± 233 ms

6.1.3.4 Influences of the force level and activation velocity on onset latency

The force level had no influence on latency (Table 6.3). Trends towards less variance of latency and less exclusion of trials with higher activity level were noted (Table 6.1and Figure 6.4). No correlation between activation velocity and onset latency was detected, r = -.02.

Table 6.3. Latencies between EMG and US onsets of gluteus medius activity and results of ANOVA for the influence of the activation level; ANOVA was performed alternatively for 40% - 80% levels because of the high exclusion rate in the 20% level.
Huynh-Feldt correction of F statistics is reported as assumption of sphericity was not met.

Onset differences EMG - US	Mean (SD) CI, ms	SEM, ms	*MDC, ms*	Median (IQR), ms	ANOVA force level differences	ANOVA force levels without 20%
EMG – M-mode, single trials	152 (114) 130 - 174	80	*223*	166 (80)		
EMG – $P_{ulsed}W_{ave}$ Doppler, single trials	126 (119) 103 - 149	84	*234*	145 (80)		
EMG – M-mode, averaged trials	150 (79) 128 - 172	56	*155*	157 (67)	*F*=.327 *p* = 0.682	*F*=.020 *p* = 0.980
EMG – $P_{ulsed}W_{ave}$ Doppler, averaged trials	133 (108) 103 - 163	76	*211*	145 (81)	*F*=.204 *p* = 0.805	*F*=.621 *p* = 0.546

6.1.4 Discussion

This first study on onset measurements of hip abductor activity provided new insights into the muscles' timing and into US measurements of activity related muscle motion. In isometric hip abduction, activity related motion started in gluteus minimus 36 ms earlier than in gluteus medius, a finding which supports the stabilizing role of gluteus minimus.

Relative estimates of the relationship between EMG and US measured onsets of gluteus medius activity were excellent, probably due to the large range of onset values. The absolute differences between EMG and US measured onsets were highly variable with wide limits of agreement. The force level and the activation speed had no influence on the latency between EMG and motion onset.

Onset measurements by M-mode US indicated a slightly higher reliability than those by Pulsed Wave Doppler and the additional advantage of onset comparisons in different levels of depth for the same trial.

6.1.4.1 Gluteus minimus is the first activating muscle in the hip abductors

To the author's knowledge, no EMG or US investigations on the timing of hip abductor activity have been published. The finding of earlier gluteus minimus motion arose from US M-mode measurements. In the first activating muscle, the relationship between EMG and US measured onsets of muscle activity is close (Mannion et al 2008a, Vasseljen et al 2006), so that early motion onset in gluteus minimus is expected to reflect the early onset of gluteus minimus excitation. Motion onset of the later activating gluteus medius is less closely related to excitation onset (Mannion et al 2008a, Vasseljen et al 2006). Excitation may start close to or later than motion onset. A sequential onset of muscle motion is with high likelihood indicative of sequential muscle excitation. However, the time lapse between excitation of the first and the second activating muscle cannot be estimated from motion onsets.

6.1.4.2 Onset latency

In the literature, the latency between onset of electrical excitation and muscle motion is inconsistent, but markedly smaller than in Study Four. Reported onset latencies between EMG and muscle motion are in the range between 2 and 54 ms (Hug et al 2011a, Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006). These studies mostly used a data synchronization based on the ECG channel of the US system. The much larger latencies found in Study Four, 126 – 152 ms, are most likely caused by

the different synchronization technique of Study Four (section 6.2). A test on an ASPEN US system (Acuson, Mountain View, CA) indicated that the time stamp produced by the PEAK synchronization unit preceded the synchronization signals in the Aux and ECG channels by one frame (= 40 ms). The ECG synchronization signal was the latest of three tested synchronization methods (section 6.2). An early synchronization signal results in a long time span until motion onset, a late synchronization signal leads to a shortened time interval. Differences in the synchronization methods and the internal processing times of US systems may explain the inconsistent latencies between EMG and US onsets in the literature and the difference to this study (Walker et al 2002). As there is no validated synchronization standard, 'true' latency remains unclear.

6.1.4.3 Latency variation

Latency between EMG and US onset was highly variable. Standard deviations (SD) of 114 and 119 ms were documented. One likely reason for large variance is the velocity of task performance. Most literature studies had the task performed in maximal activation speed (Vasseljen et al 2009, Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006). The resulting SD of onset latencies were between 12 and 53 ms. An EMG study on the latency between armlift and abdominal muscle activity in different activation velocities documented increasing SD, up to the range found in this study, with lower activation velocity (Hodges and Richardson 1999).

A second cause for high variation is the measurement of a superficial muscle. Literature reports higher latency variation in superficial, later activating muscles than in deep, early activating muscles (Vasseljen et al 2009, Mannion et al 2008a). Probably, gluteus medius motion onset was influenced by the earlier onset of gluteus minimus motion, which increased variation.

A third reason for high variation may be visual onset detection, which resulted in less reliable onsets than a computed detection method (Vasseljen et al 2009, Vasseljen et al 2006). The SEM inflated the variation on the side of the EMG and of the US onset each by 10 – 20 ms. Also, the frame-based synchronization method inflated the variation by at least 20 ms. In conclusion, the large latency variation found in this study reflects in part realistic, clinical conditions and in part experimental imperfection that can be improved.

6.1.4.4 Motions onsets preceding excitation onset

In 7.5% of M-mode and 9.5% of Pulsed Wave Doppler trials muscle motion onset was registered before EMG onset, a finding shared by all studies that compared EMG and US onsets (Vasseljen et al 2009, Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006). Earlier gluteus minimus motion onset and a lateral transmission of gluteus minimus motion towards gluteus medius, e.g. by fascial attachments between both muscles (Al-Hayani 2009, Pickard 2005, pp. 45/46), is the most likely mechanism for motion onsets preceding electrical excitation. The relevance of the transmission of mechanical energy by connective tissue is a relatively new area in the understanding of muscle physiology (Gillies and Lieber 2011, Huijing 2009, Sheard and Duxson 2002, Huijing and Baan 2001). Whether this early motion plays a meaningful role in synergistic activity, e.g. adjustment of filament distance or facilitation of excitation is yet unclear.

6.1.4.5 Comparison of M-mode and Pulsed Wave Doppler measurements

M-mode onsets were slightly more reliable and concordant with EMG onsets. A higher temporal acuity of M-mode has been stated (Feigenbaum 2010, Anderson 2007, p. 78). On the other hand, two studies that compared M-mode to Doppler measurements reported a comparable detection of motion (Westad et al 2010, Fleming et al 1996). The limitations of temporal acuity by the synchronization method did not allow for detecting small differences between the modes. Whether M-mode onset measurements are more accurate and whether a difference between modes is clinically relevant, warrant further examination.

The study indicated that the M-mode presentation of activity offers advantages. The M-mode trace allows for the distinction and comparison of motion in different levels of depth for the same trial. A comparison of motion in adjacent muscles is valuable for assessing motor control, and is not feasible using Pulsed Wave Doppler.

The detection of muscle motion by M-mode is based on greyscale values, echo amplitude. Motion detection by Pulsed Wave Doppler is based on spectral changes, echo frequency. The question whether M-mode or Pulsed Wave Doppler measure onset more accurately implies the question for the more sensitive technology to detect motion. A pennate muscle moves in three dimensions during activity. The muscle thickens, shortens and changes the fascicle angle. The sensitivity of Doppler based modes for motion is strongly angle dependant. The sensitivity for the detection of each motion component differs in Doppler, a source of artefacts (Lindberg et al 2011, Hoskins

et al 2010, pp. 114-15) and a limitation in the application of Doppler US for measurements of muscle motion (Vasseljen et al 2009). The influence of the scanning angle on motion detection by M-mode US is currently undetermined. As the resolution of an US transducer is highest along the sound beam and worst in the elevation plane, it can be assumed that motion detection is most sensitive along the US beam and least in the elevation plane (Hoskins et al 2010, pp. 64-66), however with smaller differences than Doppler detection. In summary, onset measurements of muscle activity should be taken by M-mode US.

6.1.4.6 Limitations

The major technical limitation of Study Four, the video frame-based synchronization technique has been discussed already in its influence on the study results. The exclusion rate in Study Four was 25%. Most of the excluded trials were in the 20% MVIC level and did not demonstrate a definite rise of the EMG signal. As the M-mode US demonstrated motion in deeper muscle levels, it is likely that isometric abduction in low force can be performed without activating the superficial gluteus medius muscle. Thus, the high exclusion due to missing EMG onsets in the 20% MVIC level further supports the use of M-mode US. The high latency variation between EMG and motion onsets, which has been discussed in section 6.1.4.3, lead to an underpowered ANOVA for the influence of the force level. However, no trends for an influence of the force level on onset latency were detectable.

This study was limited to the comparison of M-mode to Pulsed Wave Doppler US. An inclusion of M-mode based Colour Doppler, TDI, would have extended the comparison including a Doppler technique with a larger sampling area and existing applications in motor control studies (Mannion et al 2008a, Pulkovski et al 2008). An existing TDI study noted that in spite of the larger sampling area, TDI does not differentiate between muscle regions to allow for a comparison between adjacent muscles (Mannion et al 2008a) and does therefore not overcome the main limitation of Pulsed Wave Doppler.

6.1.4.7 Conclusion

M-mode US indicated a sequential onset of activity related gluteus minimus and gluteus medius motion, an innovative insight into mechanical patterns of muscle

activity. Study Four confirmed a complex in-vivo relationship between electrical excitation and motion of muscle, influenced by yet insufficiently determined other factors.

6.1.5 Acknowledgements

The loan of the ANTARES and Aspen US systems by Siemens Medical Systems, Western Australia, and the great support by Paul Davey is thankfully acknowledged.

Study Four resulted in unexpectedly large latencies between EMG and US onset. The synchronization method used in Setup One was originally designed for kinematic studies and differed from the synchronization method used in comparable studies (Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006). Uncertainties in the synchronization of US and externally recorded data have been reported (Walker et al 2002). It was hypothesised that the long latencies measured in Study Four were confounded by the synchronization method. The following section presents an experiment in which three different synchronization methods were compared.

6.2 Validity of synchronization of ultrasound imaging with external data needs reassurance[6]

6.2.1 Introduction

Synchronization of M-mode US imaging with external data is required for comparing US measurements to an external event, such as the onset of electrical excitation in measurements of the electromechanical delay. Usually, an electrically induced synchronization signal is displayed using the ECG channel (Chen *et al.*, 2009; Pulkovski *et al.*, 2008) (Figure 6.5). System-internal ECG signal processing may lead to a delayed delineation of the synchronization signals. Delayed synchronization and the lack of an absolute temporal reference threaten valid data comparisons in the time domain (Walker *et al.*, 2002).

[6] This experiment has been submitted for publication as a Short Communication with the co-authors Paul Davey and Janice McKay.

6.2.2 Experiment

Three options to produce a synchronization signal on an M-mode US trace of an ASPEN™ US system (Acuson, Mountain View, CA; L7 linear transducer, M-mode highest sweep speed) were tested. A split switch signal was send into (1) AUX entrance, (2) ECG entrance of the US system, and (3) entrance of an event synchronization unit (PEAK Performance Technologies Inc., Centennial, CO, USA), to produce a time stamp on the concurrent video frame (US data recorded on VCR, 25 frames per second) (Figure 6.5).

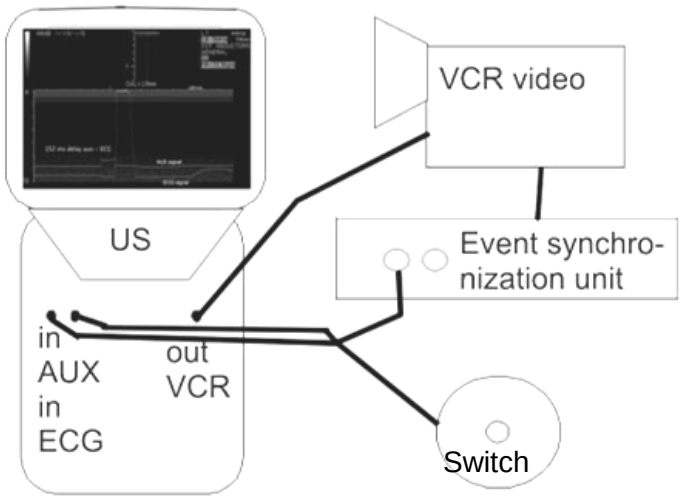


Figure 6.5. Experimental setup for the test of three different methods of ultrasound data synchronization.

6.2.3 Result

The time stamp produced by the PEAK synchronization unit appeared first, one frame (= 40 ms) before indication in the AUX / ECG trace. In the AUX trace, the switch signal was indicated by a clear signal rise, the ECG trace showed only a small shift not detectable with a small signal size. The main peak in the ECG trace occurred approximately 150 ms later (slight differences between trials) (Figure 6.6).

6.2.4 Discussion and conclusion

The synchronization signals from the three synchronization methods were demonstrated with an interval of 192 ms between the first and last appearance of

synchronization signals. An electromechanical delay measured with the PEAK synchronization would be considerably longer than one measured using ECG synchronization. Synchronization using video recording lacks temporal resolution due to the low video frame rate.

More current US systems can be expected to indicate less internal delays due to higher computing capacities. Split signal tests in a current Toshiba Xario XG (SSA-680A, Toshiba Pty Ltd, North Ryde, NSW, Australia) revealed delays of 0 to 2 ms; 9 ms delay could be provoked by rapid signal sequence. Comparisons of AUX and ECG synchronization are less relevant in current US systems, but may have large influence when working with an older US system or for comparison to studies on older US systems.

Synchronization introduces uncertainty in measurements between US and external data. Currently, the AUX channel appears to provide least internal delay, although a validation against an absolute reference is lacking. The ECG channel should be tested against the AUX channel to discover potential delays by ECG signal processing.

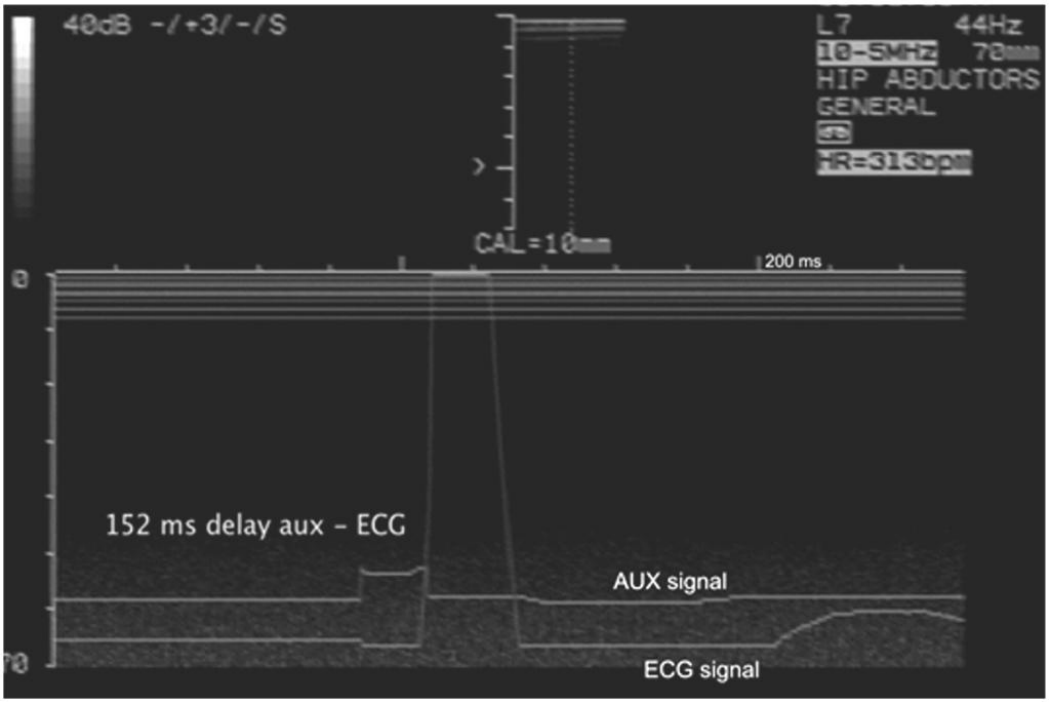


Figure 6.6. On bottom of the M-mode image two signal traces of a split switch signal entering the ultrasound system by the AUX and ECG entrances, ECG signal maximally enlarged. Note the delay of the main spike in the ECG trace.

6.3 Onset measurements by M-mode ultrasound: which scanning plane?

The data of Setup One included a series of abduction trials in M-mode in the transverse scanning plane. A study was undertaken with the objective to compare onset measurements in the longitudinal and the transverse scanning planes by reliability and agreement with EMG onset. This study did not reveal differences between the scanning planes. However, due to the inaccurate synchronization technique, the data were not appropriate for detecting small differences. Therefore, the study is not presented.

Further, the time lag between gluteus medius and gluteus minimus onset was compared between the planes. This comparison was limited to the M-mode data and not affected by synchronisation accuracy. Again, no difference between the planes could be detected. No literature was found which has examined differences in motion detection by M-mode US between the scanning planes. Currently, there is no evidence that motion detection by M-mode US differs between the scanning planes.

As Chapter Four, section 4.2 indicated a more stable transducer position in longitudinal scanning, the longitudinal plane was preferred for scanning in Setup Two in which onset measurements by M-mode US were further investigated.

A major issue in motion measurements by US is the distinction of activity related motion from other motion. The following section provides a documentation of M-mode observed motion that does not originate from muscle activity.

6.4 Other sources of motion than muscle activity in M-mode US

In Chapter Four, section 4.1.4.3 and Figure 4.3 M-mode observed motion elicited by a pulsating vessel has been demonstrated. Figure 6.7 provides an example in which motion originating in activity onset comes into motion from vessel pulsation. Comparable to the ECG artefact in EMG, vessel pulsation may obscure the onset of motion from activity.

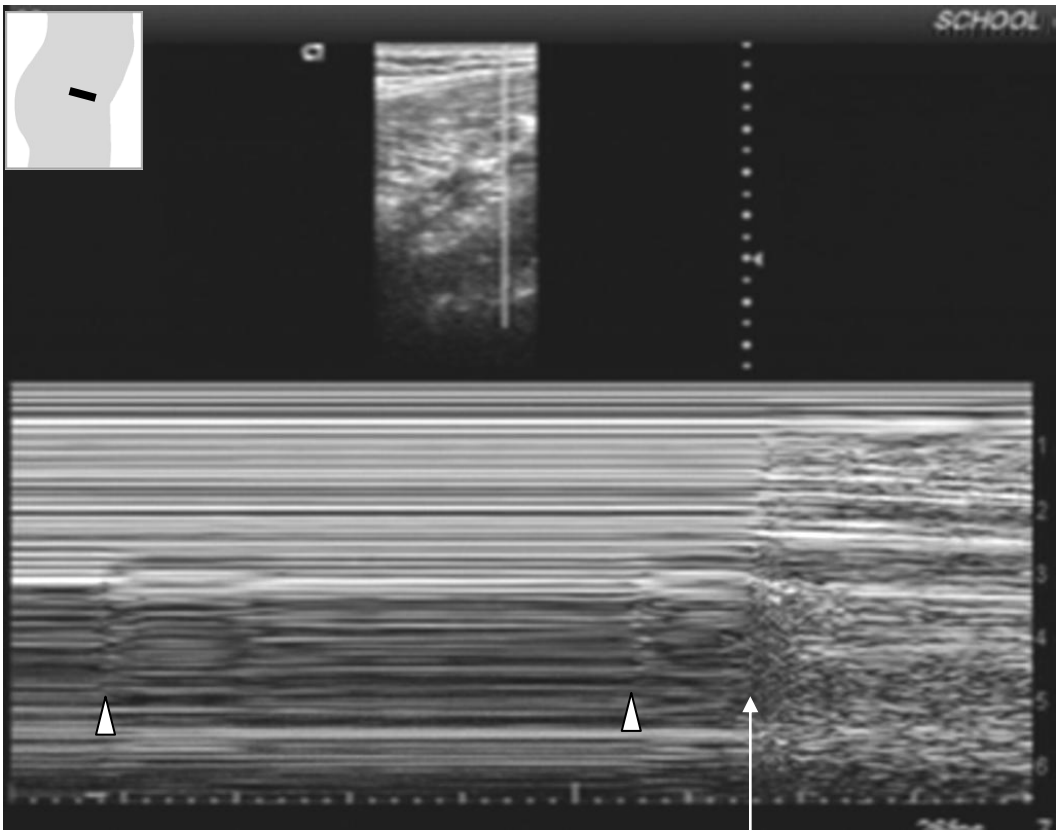


Figure 6.7. Regular motion by vessel pulsation (arrowheads). Motion by activity onset (arrow) adds into motion by pulsation.

Motion of the US transducer may create similar patterns as motion of muscle. As an experiment, the US transducer was manipulated into (1) tilting, (2) rocking and (3) gliding motion (Chapter Four, section 4.2.4 and Ophir et al 1999) on the in 30% MVIC sustained, activated hip abductors. Figures 6.8 – 6.11 document examples of M-mode traces including the three types of transducer manipulations.

Transducer induced motion is characterized by a synchronous pattern of slight motion over the full depth of the image. Transducer induced motion may be

misinterpreted as slight muscle motion. Careful control of transducer motion is necessary to avoid a mixture of different sources of motion.

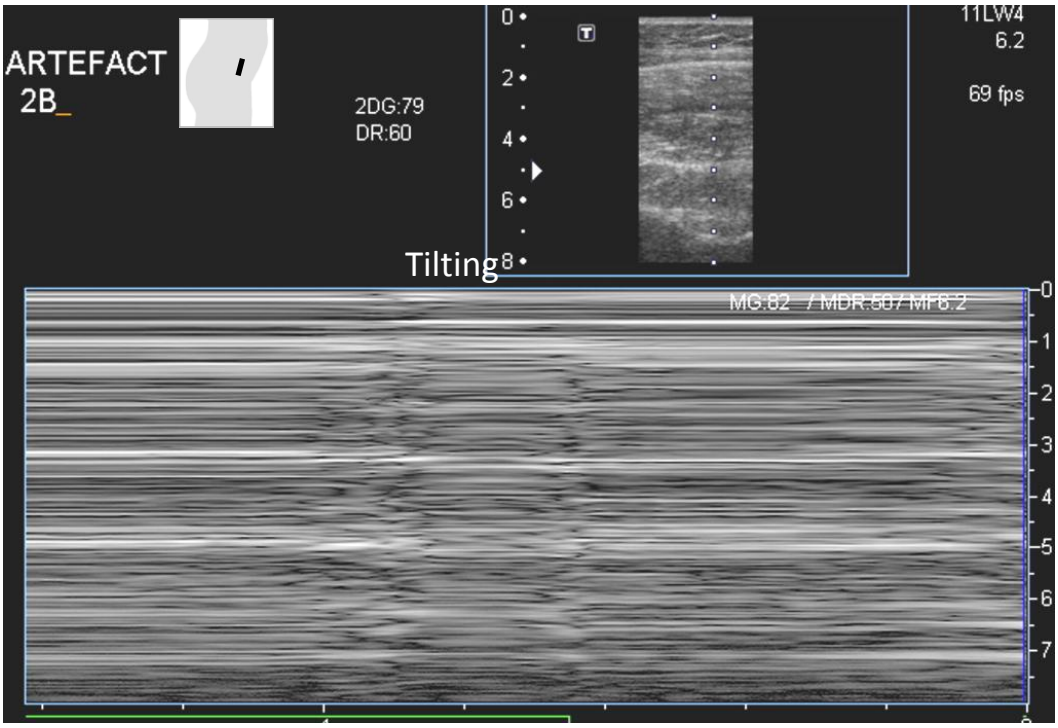


Figure 6.8. Transducer tilt on 30% MVIC sustained activated hip abductors, subject 1.

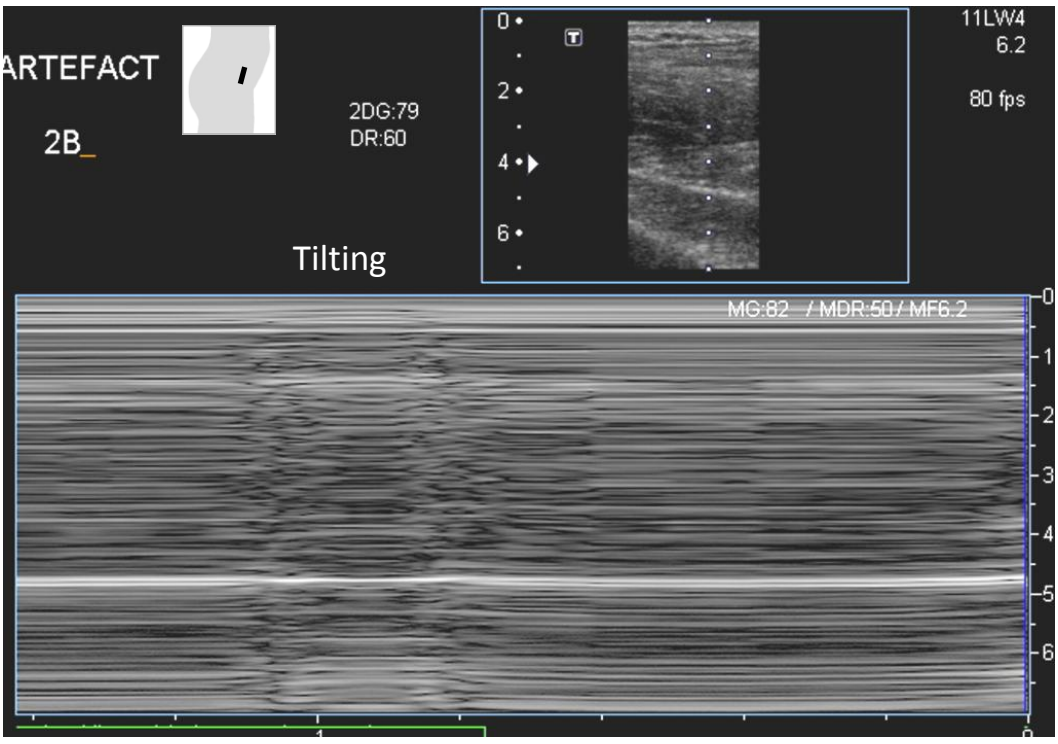


Figure 6.9.Transducer tilt on 30% MVIC sustained activated hip abductors, subject 2.

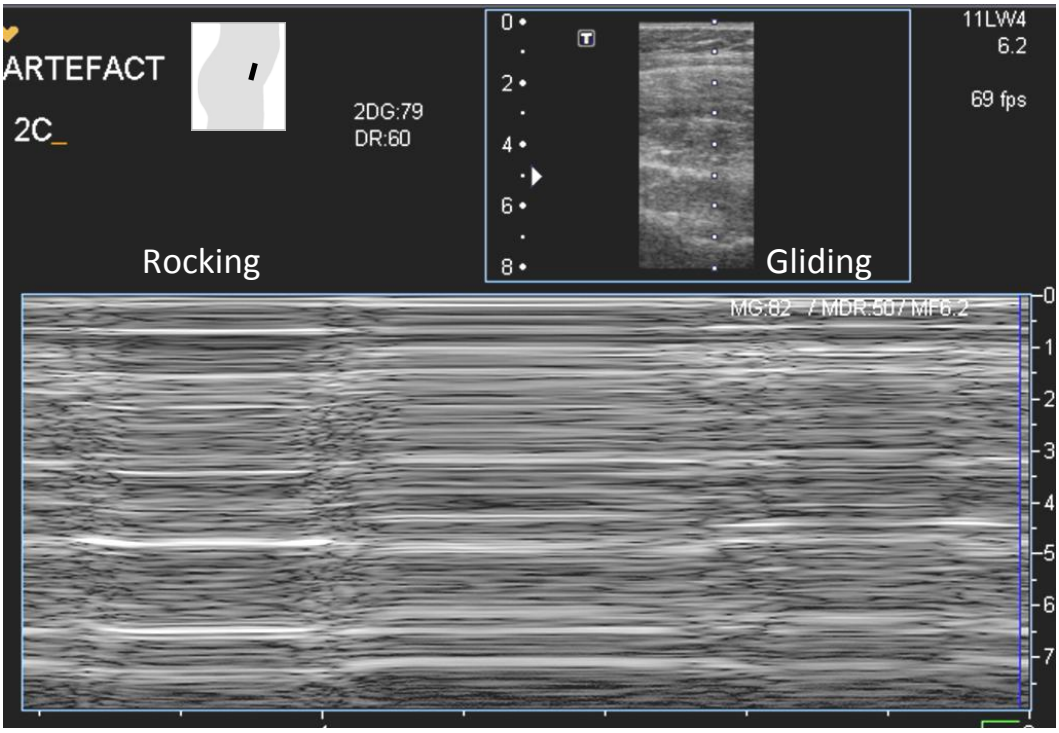


Figure 6.10. Transducer rocking and gliding on 30% MVIC activated hip abductors, subject 1.

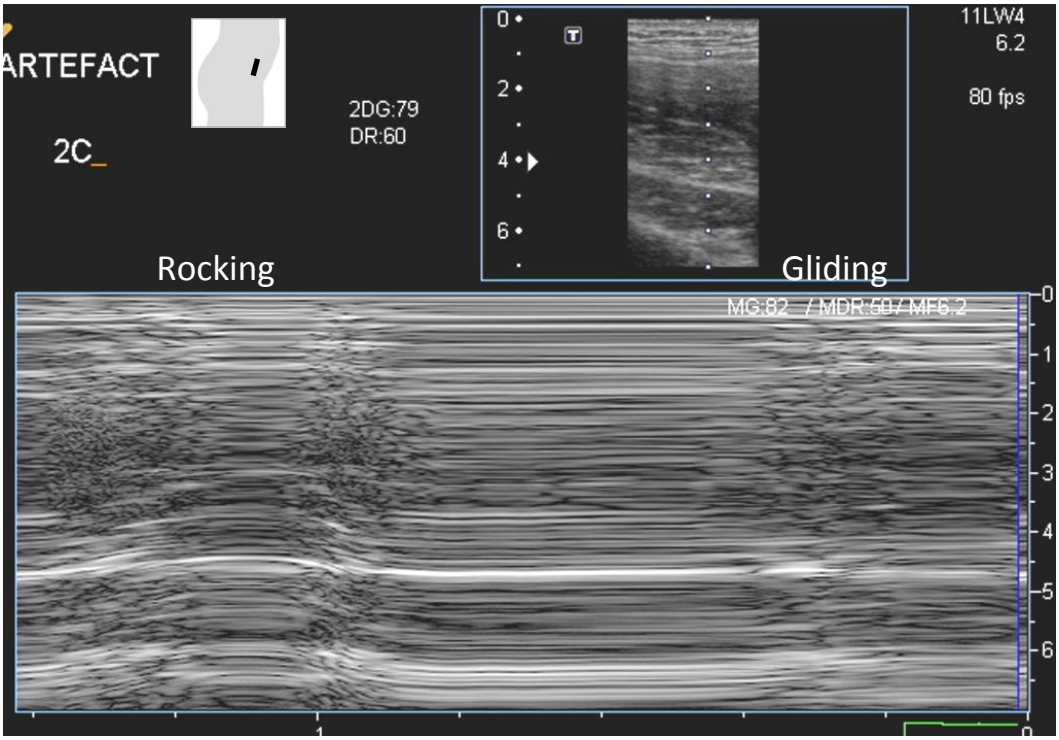


Figure 6.11. Transducer rocking and gliding on 30% MVIC activated hip abductors, subject 2.

The previous documentation relates to the motion induced by other sources than muscle activity. The following example documents a motion artefact that has been observed in both research Setups. M-mode patterns of motion may be delineated deep to the iliac periosteum (Figures 6.12 and 7.1) in the iliac bone. It has to be assumed that these observations of motion originate in mirror-image or reverberation artefacts (Hoskins et al 2010, pp. 73/74) based on sound-reflection by the iliac periosteum.

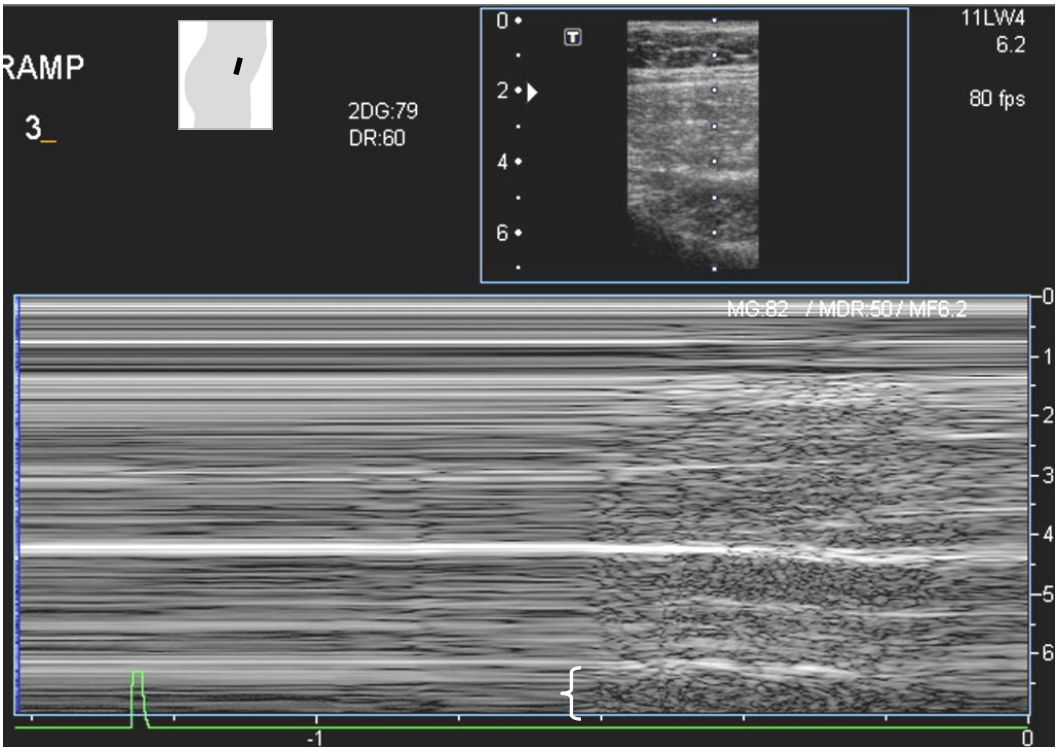


Figure 6.12. Regularly, an artefactual appearance of motion deep of the iliac periosteum (bracket) could be observed in both Setups.

6.5 Summary of the findings of Chapter Six for US measurements of onset of activity related motion

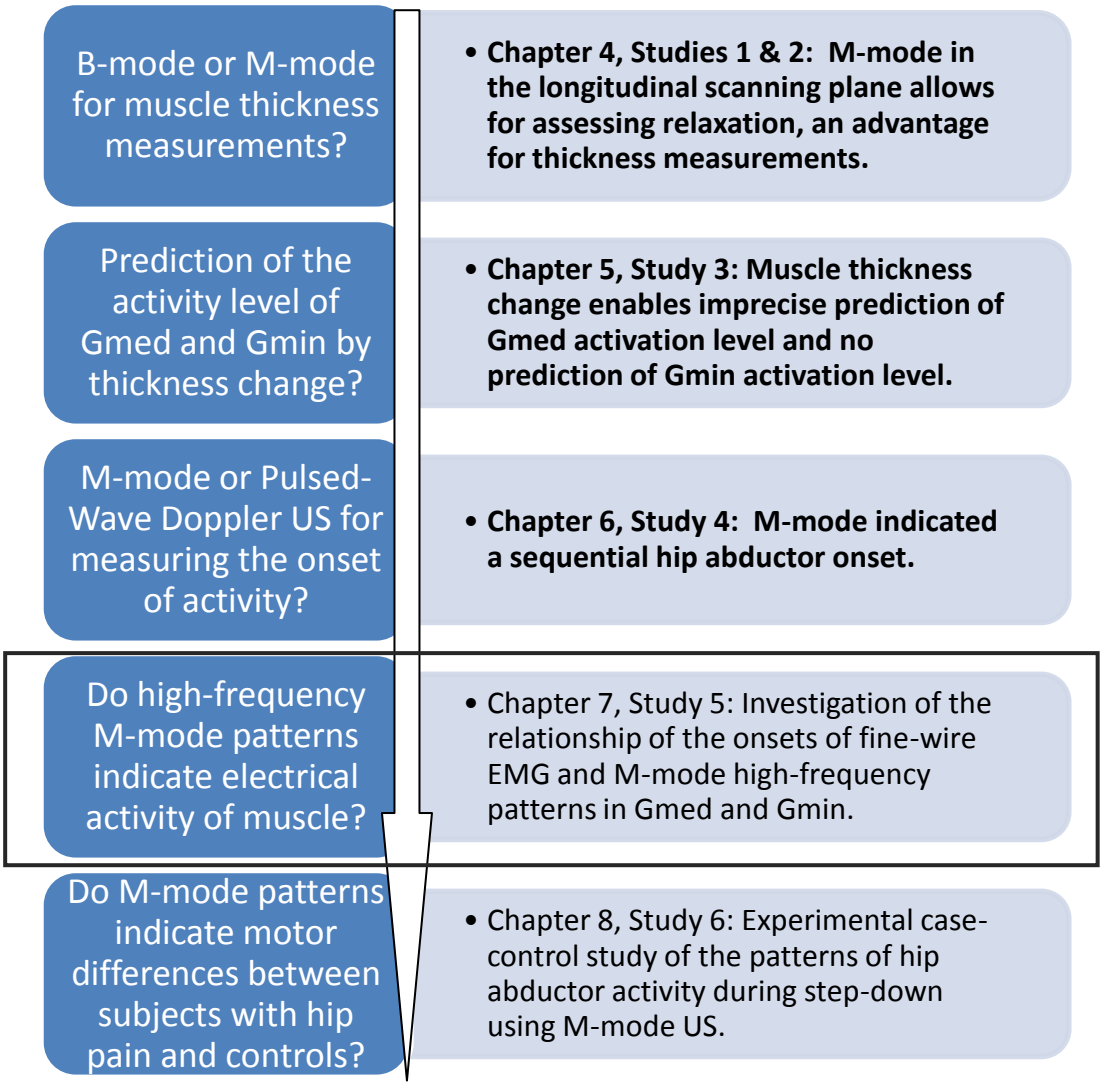
M-mode US enables the direct view on a mechanical aspect of muscle activity. It reveals differences between superficial and deep muscle activity that are not detectable by surface EMG.

The main argument for the use of M-mode US for onset measurements is the option to compare motion in adjacent muscles in the same trial. The main argument against the use of Doppler US is the angle dependency of motion detection in Doppler modes.

The high temporal resolution of M-mode US traces needs to be preserved in digital data storage. Optimal synchronization of US and external data needs further research. Currently, the AUX entrance offers an option less affected by system internal signal processing than the ECG channel.

The onsets of electrical excitation and muscle motion are closely related but their exact relationship not predictable. Motion onset within muscle may be induced by electrical excitation or by motion from other sources, such as activity of an adjacent muscle or a pulsating vessel. Setting up US onset measurements, a careful control of potential sources of motion is necessary.

The M-mode trace of muscle activity indicates different greyscale patterns during the initial phase of activity. Existing studies examined only the first onset of greyscale indicated motion. The following chapter introduces the development and the application of further, innovative motion variables that are based on M-mode greyscale patterns.


B-mode or M-mode for muscle thickness measurements?
• Chapter 4, Studies 1 & 2: M-mode in the longitudinal scanning plane allows for assessing relaxation, an advantage for thickness measurements.
Prediction of the activity level of Gmed and Gmin by thickness change?
• Chapter 5, Study 3: Muscle thickness change enables imprecise prediction of Gmed activation level and no prediction of Gmin activation level.
M-mode or Pulsed-Wave Doppler US for measuring the onset of activity?
• Chapter 6, Study 4: M-mode indicated a sequential hip abductor onset.
Do high-frequency M-mode patterns indicate electrical activity of muscle?
• Chapter 7, Study 5: Investigation of the relationship of the onsets of fine-wire EMG and M-mode high-frequency patterns in Gmed and Gmin.
Do M-mode patterns indicate motor differences between subjects with hip pain and controls?
• Chapter 8, Study 6: Experimental case-control study of the patterns of hip abductor activity during step-down using M-mode US.

7 Chapter Seven: M-Mode High-Frequency Patterns to Assess Deep Muscle Activity

Overview of Chapter Seven

- Section 7.1 introduces a new type of M-mode motion variable, high-frequency patterns.
- Section 7.2 discusses the computed detection of M-mode high-frequency onsets and introduces the Teager-Kaiser Energy Operator.
- Section 7.3 presents a method for the computed onset detection of high-frequency onsets in M-mode US traces.
- Section 7.4 examines the influence of muscle fibre size on the computed onset detection of M-mode high-frequency patterns.
- Section 7.5 reports Study Five in which the onsets of electrical excitation of the deep gluteus medius and minimus muscles were compared to the new M-mode variable of muscle motion, the onset of a high-frequency pattern.
- Section 7.6 documents muscle-specific challenges of fine-wire electrode insertion into the gluteus medius and minimus muscles.
- Section 7.7summarizes the research findings on onset detection by M-mode US.

7.1 M-mode measurement variables of muscle motion

7.1.1 Introduction

Muscle activation starts with electrical excitation that triggers a muscle contraction following a short time span of electromechanical delay. Muscle contraction becomes visible as muscle motion by M-mode US (Hug et al 2009). The sequence of electrical and mechanical events in the activation process is referred to as excitation-contraction coupling (Lieber 2010, pp. 41-45). Existing studies that compared the onset of EMG measured electrical excitation with the US measured onset of muscle motion (Vasseljen et al 2009, Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006) and also Study Four (Chapter Six, section 6.1) suggest that some muscle motion occurs independently from local excitation (Vasseljen et al 2009, Mannion et al 2008a, Vasseljen et al 2006). The interpretation of muscle motion that is not accompanied by electrical excitation is unclear.

US measurements that are indicative of the electrical excitation of muscle have been sought. Strain Rate US has been used to detect a thickening motion of muscle which proved to be a superior indication of electrical excitation (Vasseljen et al 2009). Strain Rate US is a Doppler based mode that indicates an extending or compressing movement by comparing the velocities of two points along the sound beam and dividing their difference by their distance (Vasseljen et al 2009). Strain Rate US identifies the thickening motion of a muscle only when the muscle thickens along the US beam. Due to this limitation, 50 % of the subjects of the study of Vasseljen et al (2009) could not be included in the measurements. The Strain Rate US studies (Westad et al 2010, Vasseljen et al 2009) indicated that specific motion variables may be indicative of electrical muscle excitation. Under the premise that muscle excitation is the most relevant phenomenon for assessing muscle activity, research for a broadly applicable US measurement that is indicative of muscle excitation is continued.

M-mode traces of isometric abductor activation demonstrate a sequence of different greyscale patterns in the course of the activation process (Figure 7.1), suggesting the development of different M-mode variables of muscle motion.

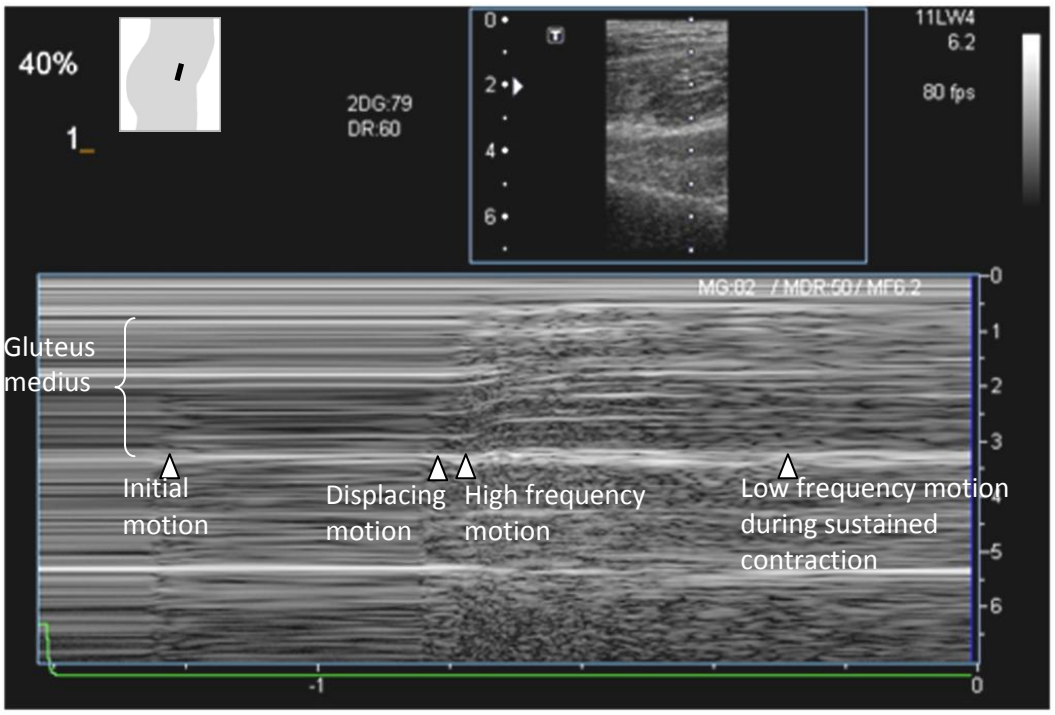


Figure 7.1. M-mode US demonstrates different greyscale patterns in the course of activation, in this example indicated at the trace of deep gluteus medius motion.

7.1.2 A suggested explanation of the formation of M-mode greyscale patterns during muscle activity

The fascicular structure of muscle tissue (Figure 7.2) may be portrayed as sequence of black and white stripes (Figure 7.3). The US beam provides a one-dimensional 'tunnel view' through muscle tissue (Figure 7.4). The US beam captures hyperechoic interfaces separated by hypoechoic zones throughout the muscle (Figure 7.4). In relaxation, the same interfaces stay in the focus of the sound beam without a change in greyscale or position in the M-mode trace. Slight motion changes the angle of reflection of interfaces and induces a change in greyscale in the M-mode trace (Figure 7.1 and Figure 4.3, motion transmitted by the pulsation of a vessel). More intense muscle motion transfers an adjacent interface into the focus of the sound beam, a displacement recognizable by a disruption of M-mode lines and a change from bright to dark or vice versa. More continuous motion causes the detection of a different interface every few milliseconds, a pattern of black and white change. The more rapid the motion, the higher is the frequency of black and white change. The frequency of black

and white change is an indicator of the speed with which muscle tissue moves through the sound beam.

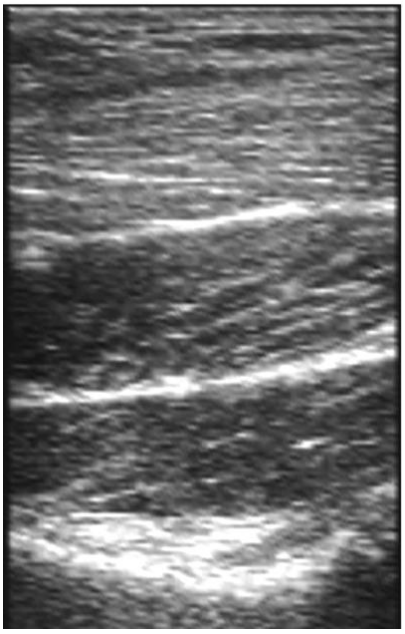


Figure 7.2. B-mode US image of the fascicular structure of muscle tissue.

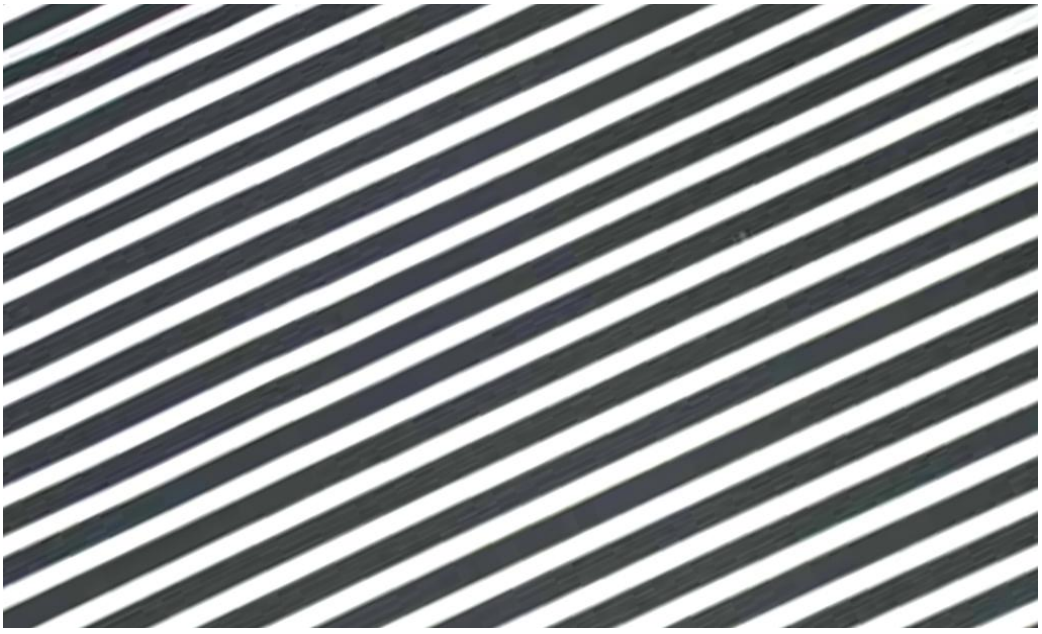


Figure 7.3. The US appearance of the fascicular structure of muscle tissue may be modelled as a pattern of black and white stripes.

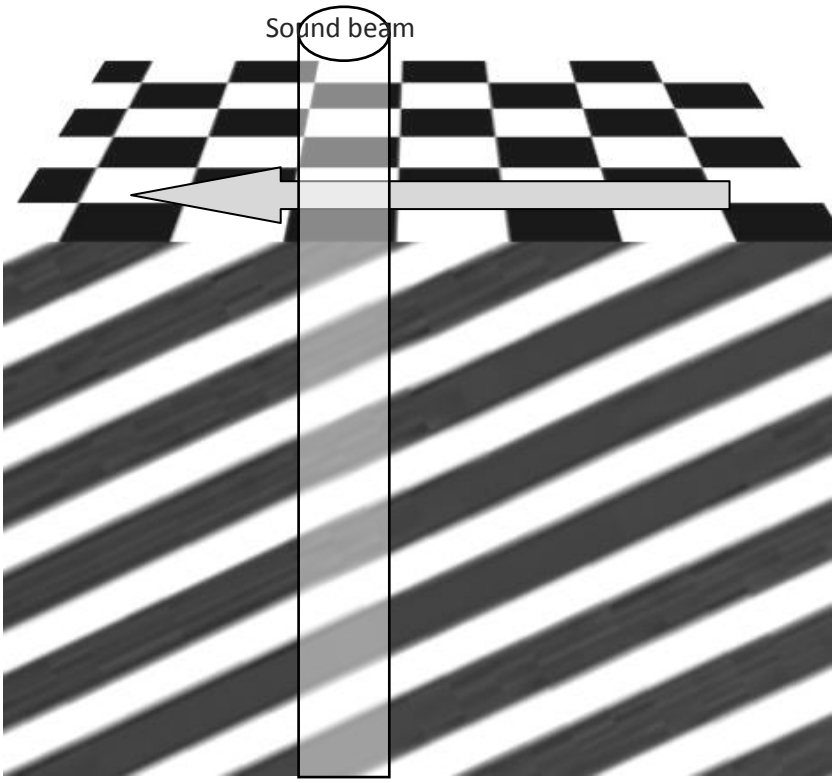


Figure 7.4. The M-mode US beam provides a one-dimensional 'tunnel' view through muscle tissue. When the muscle tissue moves (arrow) through the sound beam, a sequence of hyperechoic and hypoechoic interfaces is captured forming a pattern of black and white intervals.

It is hypothesized that high-frequency muscle motion is indicative of concurrent electrical excitation. The continued investigation of M-mode imaging of abductor activity focuses on the detection of high-frequency greyscale patterns in M-mode traces.

7.2 Computed detection of M-mode high-frequency greyscale patterns

7.2.1 Basic considerations

Different frequencies of black and white change occur during the activation process and form distinctive greyscale patterns. The visual determination of the onset of a distinct greyscale pattern is difficult and prone to bias. A research study of M-mode high-frequency patterns requires their computed detection. Computed onset detection in greyscale data is based on the translation of each pixel line into greyscale values.

Each Mode line is translated into a wave-like signal of greyscales which modulate between bright (high amplitude) and dark (low amplitude) (Figure 7.5).

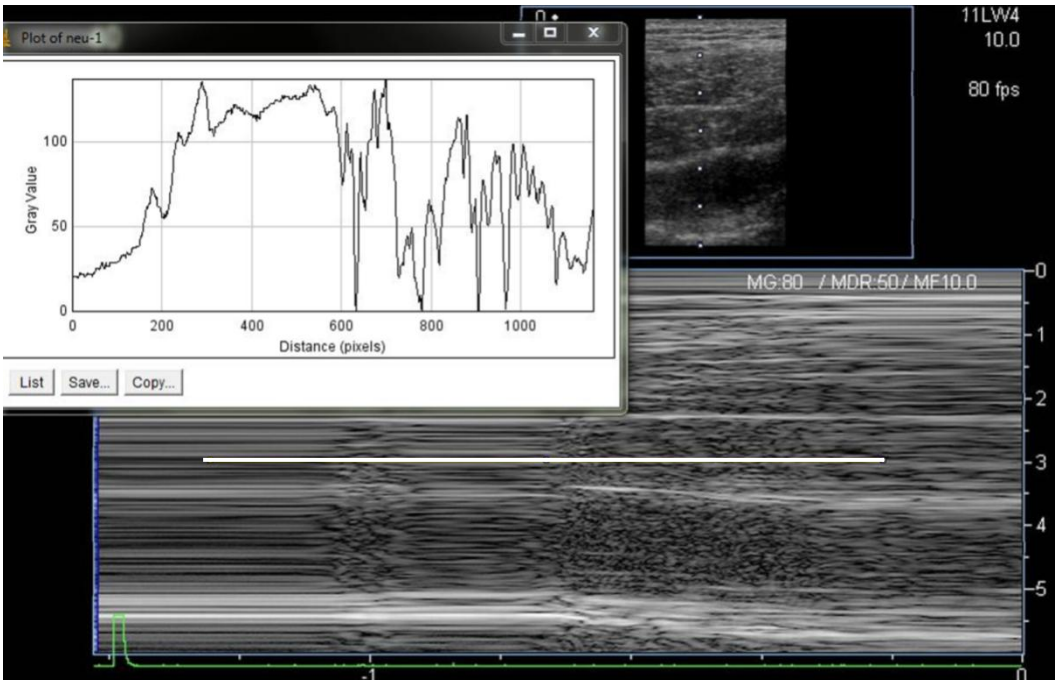


Figure 7.5. Translation of the greyscale values in the bold white line into a wave-like signal (software ImageJ, version 1.40; rsb.info.nih.gov/ij/).

Greyscale amplitude is by its absolute value meaningless for the detection of high-frequency muscle motion. The greyscale value describes the scanned tissue components. High amplitude designates hyperechoic, reflective, connective tissue; low amplitude indicates hypoechoic, contractile tissue (Walker et al 2004). It is not the absolute greyscale value but its change that indicates motion. Rapid motion is characterised by the frequency in which hyperechoic fascicular structures move through the sound beam. The computed detection of M-mode high-frequency patterns requires the computed recognition of the change of greyscale frequency.

Commonly, frequency analysis is performed by a Fast Fourier Transform (FFT). The FFT has no time domain (Figure 3.7), and is therefore not applicable for determining the instant of frequency change. Specific forms of FFT allow for an examination in the time domain, but accurate detection needs an extended sample length and the temporal resolution is limited. The determination of absolute frequencies is not necessary for onset determination in M-mode traces but the detection of the change from low to high-frequency is required. A change of signal frequency is an indication of

a change of the energy level of the signal. Energy detecting signal transformation enables the computed detection of frequency change. The Teager-Kaiser Energy Operator (TKEO) is an energy detecting signal transformation which has been used in EMG and speech analysis (Malone et al 2011, Solnik et al 2010, Lauer and Prosser 2009, Solnik et al 2008, Li et al 2007b, Li and Aruin 2005, Kaiser 1990).

7.2.2 The Teager-Kaiser Energy Operator (TKEO)

The TKEO is an algorithm that describes the energy level of a signal. The algorithm is based on a property of a regular oscillation: the energy for signal perpetuation equals the product of the squared amplitude and the squared frequency (Kaiser 1990). This relationship is the basis for the following formula (Li and Aruin 2005).

$$TKEO = x^2(n) - x(n-1) \times x(n+1)$$

Formula 7.1. The Teager-Kaiser Energy Operator, an energy detecting signal transformation.

The formula characterizes a regular relationship of adjacent points in an oscillating signal. The instantaneous energy of the signal at the point n is the difference between the squared value of n and the product of the values of the points before and after n. In a regular sine wave centred at zero the energy level is constant (Figure 7.6).

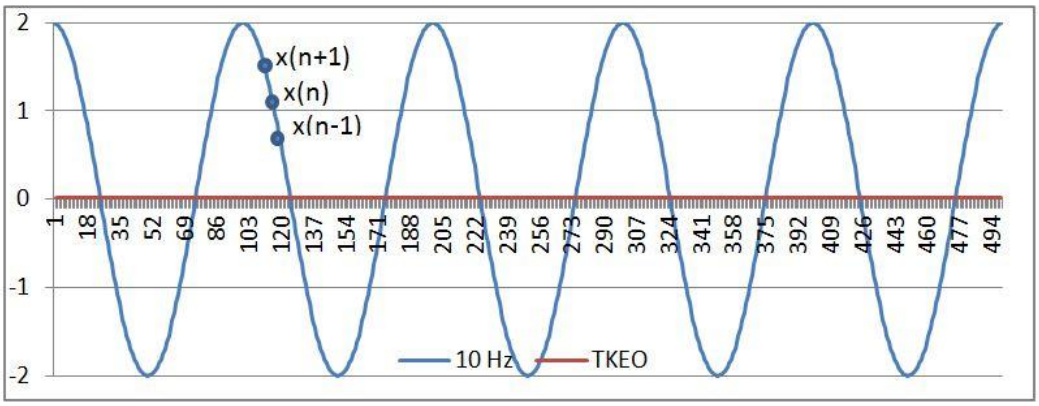


Figure 7.6. Adjacent signal points in a regular sine wave, their regular relationship is the basis of the Teager-Kaiser Energy Operator (TKEO).

To the author's knowledge, the TKEO has not been applied to US data. Basic properties of the TKEO must be explored to allow for an understanding of its

applicability on greyscale data. In the following, the TKEO is applied to a mathematically constructed wave signal. The signal was constructed using the following formula (Nelson-Wong et al 2009).

$$x_i = 2\cos\left(2\pi \frac{i}{100}\right)$$

Formula 7.2. Mathematically constructed wave signal in which the red part is determining the amplitude and the blue part is determining the frequency of the signal (Nelson-Wong et al 2009).

7.2.2.1 TKEO test one, amplitude or frequency change

A signal with the amplitude of 2 and a frequency of 10 Hz was constructed. After 500 samples, the amplitude was raised to 3 without changing the frequency. After 1000 samples and without changing the amplitude, the frequency was lowered to 7 Hz. After 1500 samples and again without changing the amplitude, the frequency was lowered to 5.5 Hz. The TKEO of the signal was processed and enlarged (*10) for better recognition. A change in the amplitude or the frequency of the wave signal induced a change in TKEO amplitude. An interruption of the regularity of the signal produced a sharp spike in the signal's TKEO (Figure 7.7).

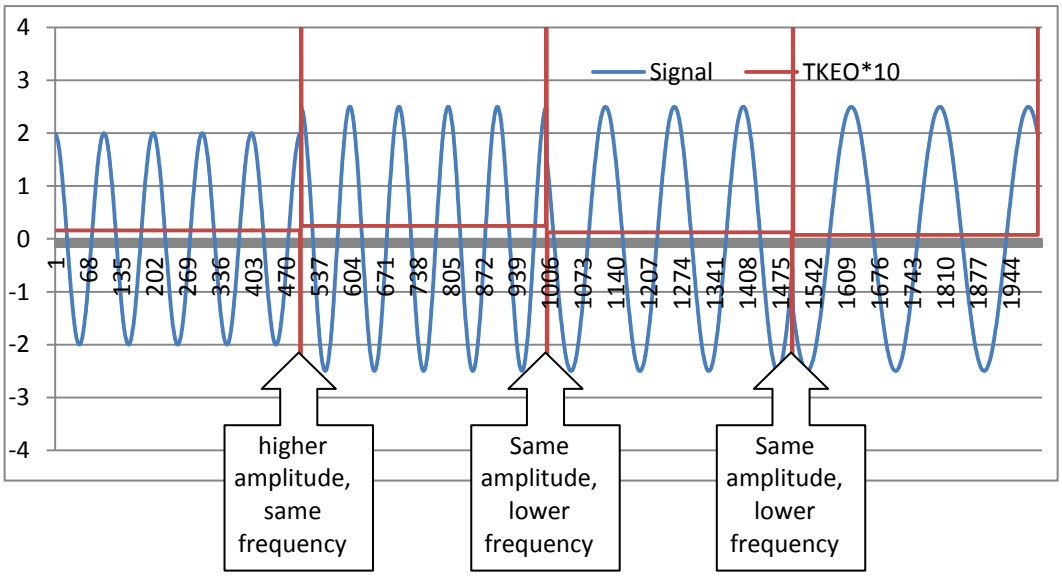


Figure 7.7. Blue: wave signal, red: TKEO*10 of the blue signal. An interruption of the regularity of the signal produces a sharp spike in the TKEO at 1000 and 1500 samples.

7.2.2.2 TKEO test two: linear increase of amplitude or frequency

A signal with a linear increase in amplitude but constant frequency was constructed (Figure 7.8). A linear increase of amplitude results in an exponential increase of the signal's TKEO transformation.

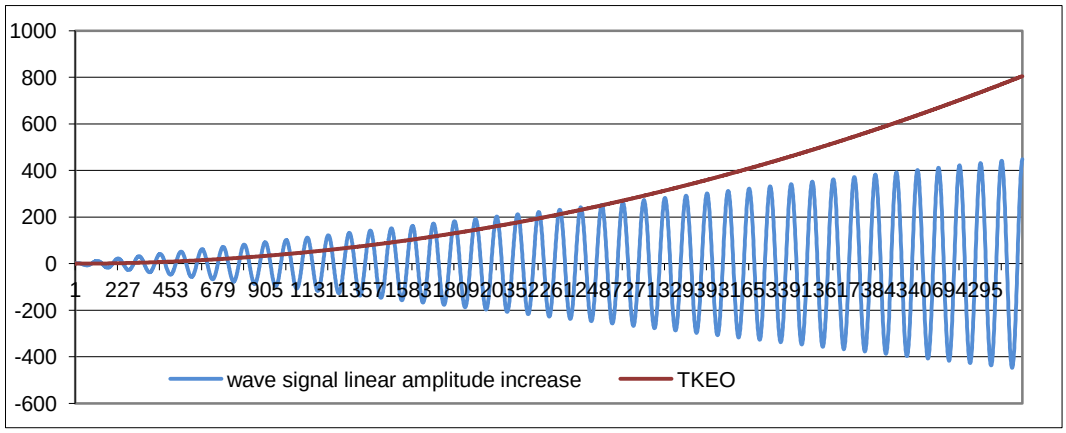


Figure 7.8. A linear increase of amplitude results in an exponential increase of the signal's TKEO transformation.

A signal with a linear increase in frequency but constant amplitude was constructed (Figure 7.9). A linear increase of frequency also results in an exponential rise of the signal's TKEO. The TKEO does not indicate whether the amplitude or the frequency of a signal changed.

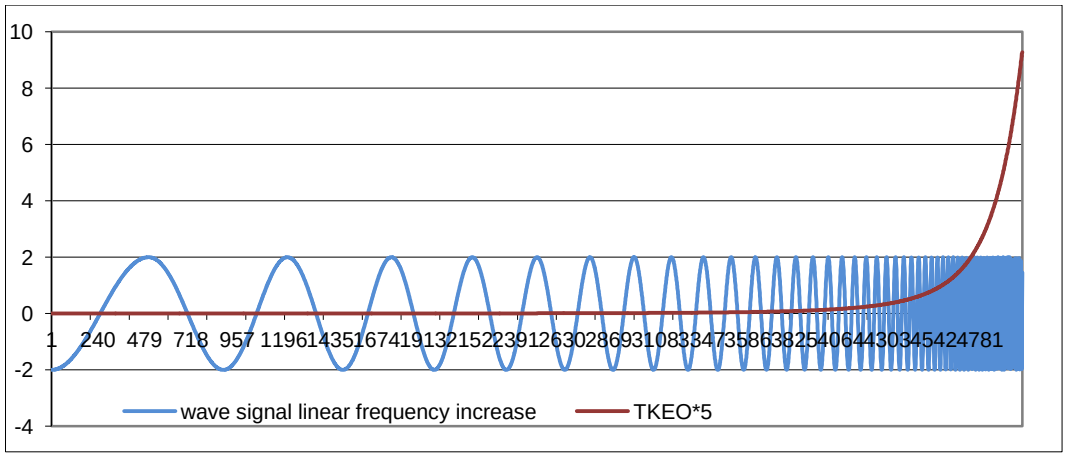


Figure 7.9. A linear increase of frequency results in an exponential rise of the signal's TKEO transformation.

7.2.2.3 TKEO Test three: wave signals that do not cross 0

Up to this point, the tests of the TKEO were limited to signals oscillating around zero. US signals consist of greyscale values that are limited to the positive range. A regular signal was transposed into the positive range to test the influence of the range on the TKEO's properties. In positive wave signals the signal's TKEO oscillates with minimal amplitude (Figure 7.10).

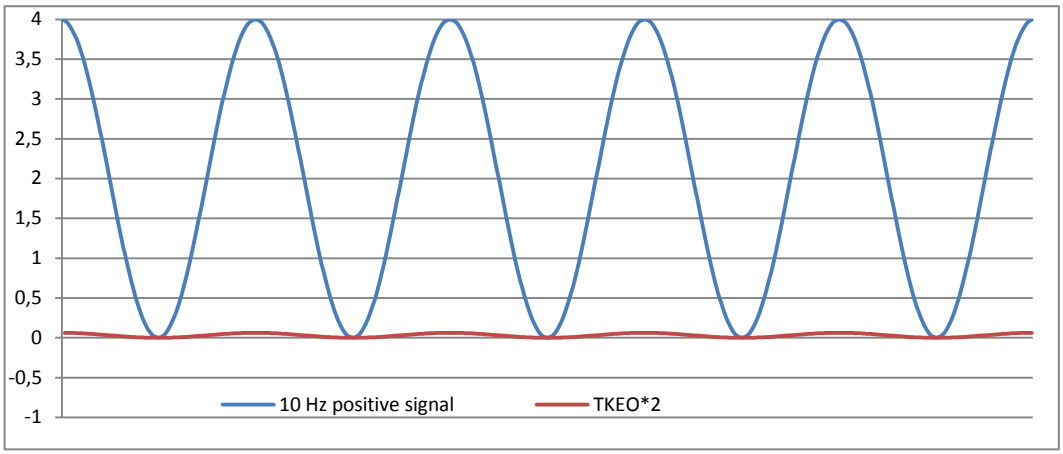


Figure 7.10. In positive wave signals the signal's TKEO transformation oscillates with minimal amplitude.

The signal of test one Figure 7.7 was transposed into the positive range. TKEO amplitude changed with amplitude and frequency change. TKEO oscillation reproduced the signal's frequency. Discontinuity of the signal induced sharp TKEO spikes in the positive and the negative ranges (Figure 7.11).

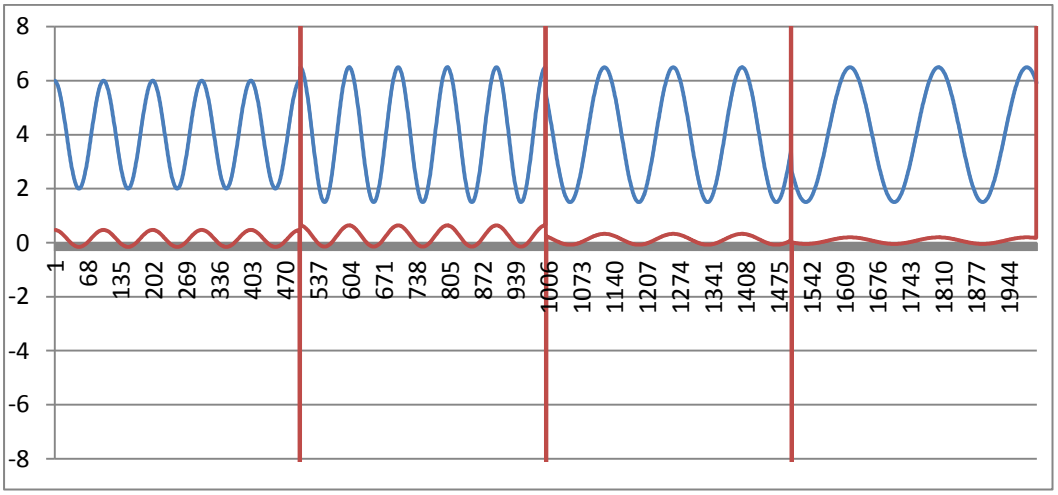


Figure 7.11. Wave signal of Figure 7.7 transposed into the positive range and the respective TKEO.

As demonstrated here, a change in the frequency of a positive wave signal is detectable using the TKEO algorithm. An application of the TKEO algorithm on US greyscale data to detect the onset of frequency change is therefore possible.

7.2.2.4 Unit of the TKEO of US greyscale data

The greyscales of an US image are descriptors of the intensity of sound reflection. The unit of intensity of US is milliwatt/cm² (Gent 1997). Theoretically and assuming a linear relationship of US intensity and greyscale translation, the unit of the TKEO would be $(mW/cm^2)^2$. In reality, most US applications use a non-linear translation of US amplitude into greyscale, so that the use of units which suggest a linear relationship is physically incorrect. In this thesis no unit is used to specify TKEO amplitude of M-mode greyscale data.

7.3 Computed detection of onset of high-frequency patterns in M-mode traces

Computed onset detection in M-mode traces was achieved using a custom-programmed LabView application (LabVIEW V8.2.1, National Instruments, Texas). The Labview application included two screens. In the first screen, the US image of the onset of hip abductor activity including the M-mode trace and the B-mode image enabled the determination of the depths of the muscles of interest. In the second screen, onset detection was performed on the M-mode trace. To start the analysis, the depth of interest and the threshold values for onset detection in the TKEO graph had to be set (Figure 7.12). The synchronization signal was detected automatically. Analysis started after the synchronization point. The program translated the greyscale values of each pixel line in the depth of interest and transformed the resulting signals by application of the TKEO. For each point in time mean and standard deviation of all TKEO lines was calculated and displayed graphically on top of the M-mode trace (Figure 7.12). Using a threshold algorithm, the instants when the standard deviation of the TKEO exceeded the threshold values of 800, 1200 and 2000 for more than 7 samples were determined as the onsets of three different intensities of muscle motion. The threshold values and the sample number were chosen by testing different options in pilot analyses.

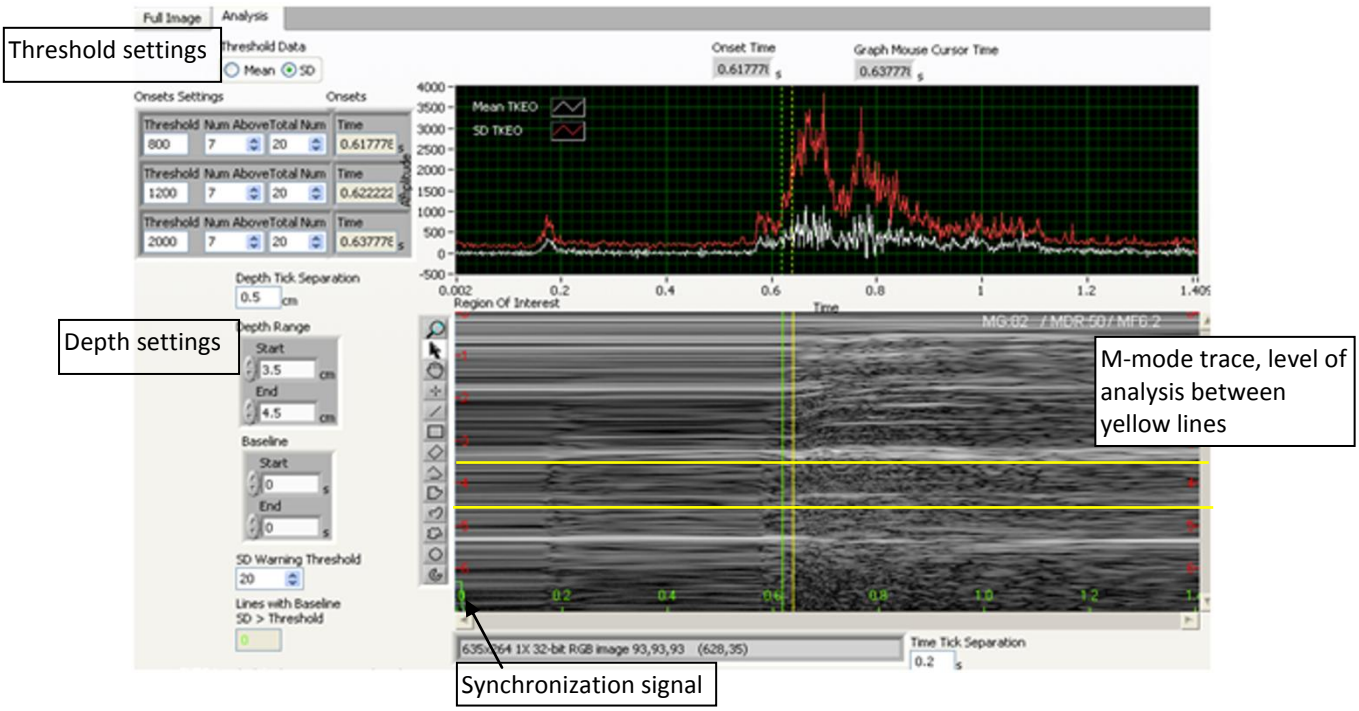


Figure 7.12. LabView screen for computed onset detection in M-mode traces, the mean (white) and the standard deviation (red) of the TKEO transformations of the pixel lines in the depth of interest (between the yellow lines) are displayed on top of the M-mode trace.

It was observed that the standard deviation of the TKEO transformed greyscale lines is more indicative of the onset of M-mode high-frequency patterns than the mean TKEO. The mean of the TKEO lines indicates the mean energy level. As energy changes occur to the positive and to the negative depending on signal amplitude of the line during relaxation (whether the line starts as a bright or a dark line), cancellation occurs by taking the average. The standard deviation of the TKEO reflects the differences in the instantaneous energy levels between lines, an information regarding the vertical dimension of the greyscale pattern and a better indicator of onset of high-frequency patterns.

7.4 The influence of fascicle size on TKEO based onset detection of M-mode high-frequency greyscale patterns

Section 7.1 explained the influence of the speed of muscle motion on the frequency of black and white change in M-mode traces using a model of black and white stripes for muscle tissue. The diameter of the muscle fascicles constitutes a

second determinant of the frequency of black and white change. Fascicle diameter depends on the muscle fibres' diameter. Fibre diameter is different between muscles (Eriksson and Thornell 1983) and muscle fibre types (Pernuš and Eržen 1994). Fascicle diameter may change with the muscle's use and the individuals age (Melichna et al 1990). A smaller fascicle diameter creates a smaller spacing between the hyperechoic stripes of the perimysium. A smaller spacing increases 'baseline' frequency. In the stripe model, fascicle diameter is the width of the stripes. The influence of fascicle diameter is examined on the mathematically constructed wave signals.

Regular wave signals (a) and (b) were constructed, (b) with double the frequency of (a), which equals half fascicle diameter. Both wave signals were manipulated to increase linearly in frequency after 1000 samples. Figure 7.13 demonstrates the two signals and their enlarged TKEO transformation. Using two as the onset threshold, onset would be detected at 2874 in signal (a) and at 1732 in signal (b).

A smaller fascicle diameter shifts TKEO onset detection towards earlier detection. Both, speed of motion (dynamic component) and fascicle diameter (structural component) influence the onset of M-mode high-frequency patterns. Both, the dynamic and the structural component are related to muscle activity.

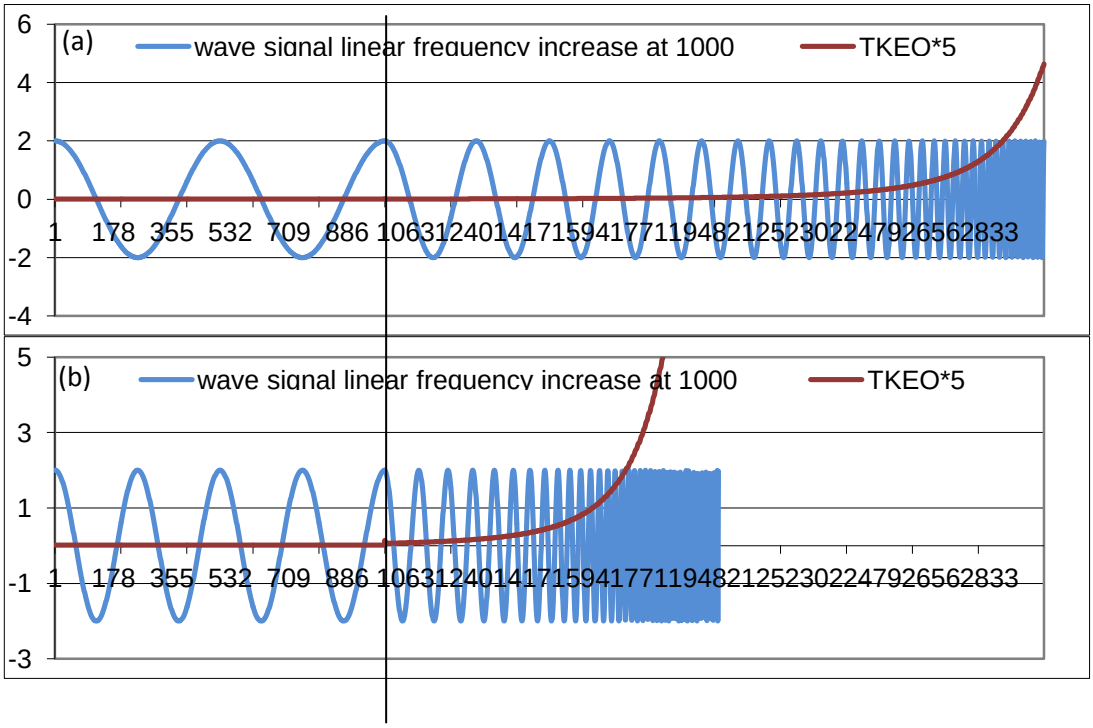


Figure 7.13 a, b. Two wave signals, (b) double frequency of (a). Both signals were manipulated to have a linear increase in frequency after 1000 samples (vertical black line, small shift of the TKEO). Using amplitude of 2 as TKEO threshold, onset from TKEO would be determined at 2874 in signal (a) and at 1732 in signal (b).

7.5 Study Five: M-mode measurements of the onset of gluteus minimus activity: are high-frequency patterns associated to muscle excitation?

7.5.1 Introduction

The use of M-mode US imaging as a substitute of invasive EMG for the assessment of the onset of deep muscle activity needs further exploration. In existing studies, only the relationship between the onset of electrical excitation and the start of muscle motion has been investigated (Vasseljen et al 2009, Vasseljen et al 2006). M-mode high-frequency greyscale patterns constitute new motion variables that are unexplored in their relationship to EMG. It is hypothesized that high-frequency greyscale patterns are an indication of electrical muscle excitation.

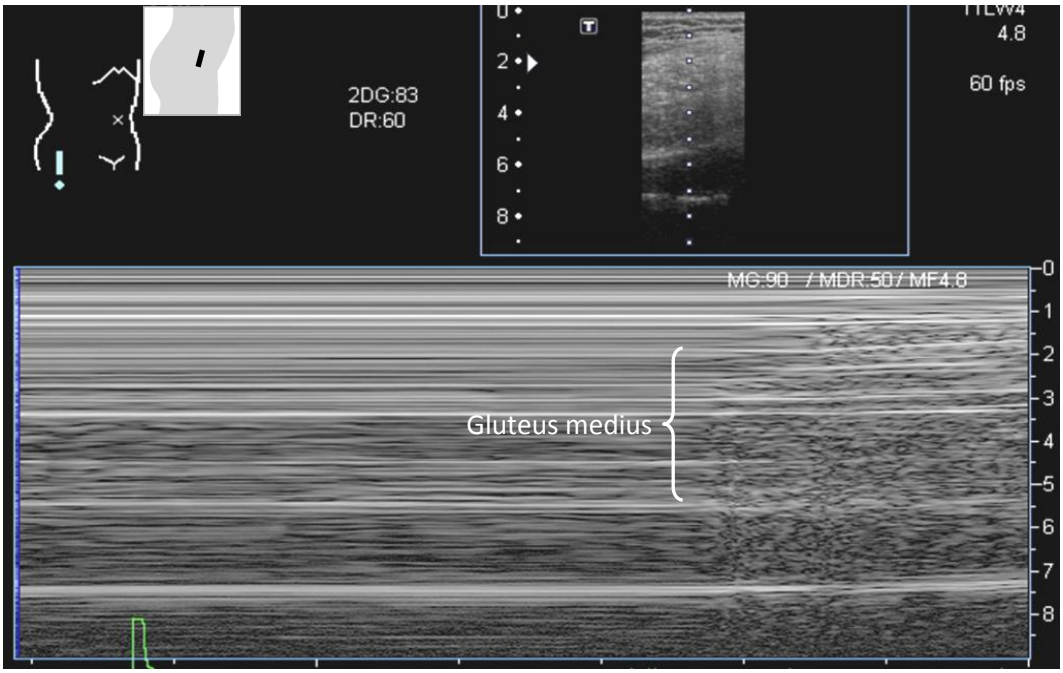


Figure 7.14. Sequential gluteus medius motion onset in 20% MVIC isometric hip abduction. Note incomplete relaxation in the deep gluteus medius and minimus muscles.

M-mode observations of sequential motion onsets within gluteus medius (Chapter Six, section 6.1.3.2) suggest a functional division of gluteus medius into a superficial and a deep part (Figures 2.7, 6.12, 7.14). The plastinated specimen in Figure 2.13 suggests that the superficial gluteus medius consists of the middle muscle part, which is thin at the anterior scanning location. The deep gluteus medius consists of the

anterior muscle part (Figure 2.13, p. 45). Surface EMG of gluteus medius represents primarily the superficial muscle but may be confounded by crosstalk of the deeper muscle part.

The objective of Study Five is to examine the relationship of the onsets of fine-wire EMG and of M-mode high-frequency greyscale patterns in the deep gluteus medius and minimus muscles.

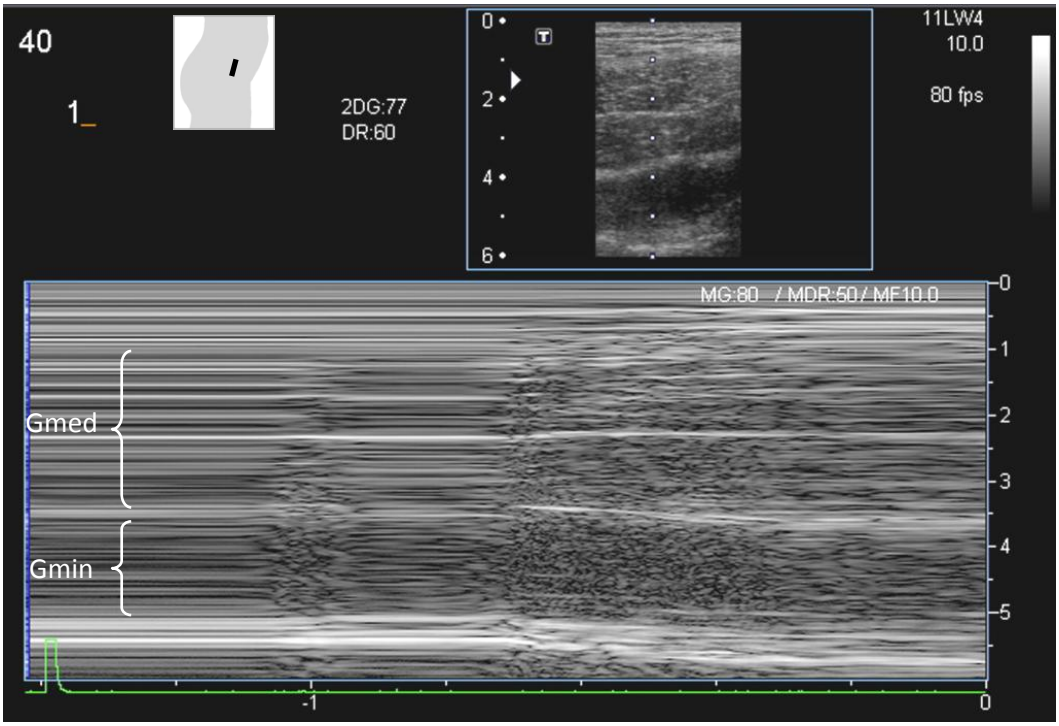


Figure 7.15. M-mode trace of gluteus medius and minimus motion during the start of isometric hip abduction in 40% MVIC. The trial demonstrates a short, pre-emptive burst of activity before full activation. Do the high-frequency greyscale patterns indicate muscle excitation?

7.5.2 Methods

Data of Setup Two were analysed in Study Five. A more detailed description of the methods of data collection is presented in Chapter Three, section 3.4.

7.5.2.1 Participants

Subjects were recruited in the School of Physiotherapy, Curtin University. Exclusion criteria were skin irritability in the hip or trunk region, hip surgery, pathology of the nervous system, medications potentially affecting the nervous system or reaction time, BMI > 32 kg/m^2. As the study was a method comparison, volunteers with hip

symptoms were included as a separate group to test whether pain may affect the agreement of M-mode US with EMG. Ethics approval was provided by the Human Research Ethics Committee of Curtin University (HR 143/2009). Subjects signed informed consent.

7.5.2.2 Procedure

Two pairs of fine-wire electrodes were inserted under US guidance obliquely into the deep gluteus medius and minimus muscles caudal and lateral to the anterior superior iliac spine (ASIS) (Figure 3.11). Free electrode ends were connected to amplifiers and secured. A surface ground electrode was positioned on the contralateral ASIS. The insertion area was covered against US gel intrusion with gauze and a transparent adhesive film (OPSite, Smith & Nephew, London, UK). The US transducer was attached to a wedge-shaped support, positioned over the fine-wire electrodes and fixed to the pelvis (Figures 3.14 and 3.15).

Isometric, right hip abduction was recorded in supine and from neutral joint position against a dynamometer. The right shank was housed in a splint to control for leg rotation (Figures 3.10 and 3.12). The level of maximal isometric abduction force (MVIC) was determined in three ramp activations. With visual feedback of actually exerted and target force levels, three repetitions of 20% and 40% and two repetitions of 60% MVIC were recorded without influencing activation speed. Finally, three repetitions of 60% MVIC in maximal activation speed were requested.

7.5.2.3 Instrumentation

7.5.2.3.1 Torque

Data of torque onset and level were collected at 2000 Hz using a Mecmesin dynamometer (AFG-500N, Slinfold, UK). Torque and EMG signals were recorded and displayed by Vicon Nexus software (Vicon Motion Systems Ltd, Oxford, UK). Torque signals were low-pass filtered with a zero lag 4^{th} order Butterworth filter at 40 Hz.

7.5.2.3.2 Fine-wire electromyography (in detail in section 3.4.6.2)

Fine-wire EMG was recorded on a Octopus AMT-8 EMG system (Bortec Electronics Inc., Calgary, Canada), using a sampling rate of 2000 Hz. Signals were demeaned, zero-lag band-pass filtered (10 - 900 Hz) and full-wave rectified.

7.5.2.3.3 M-mode ultrasound (detailed in sections 3.4.6.3 and 3.4.6.4)

A Xario XG system (SSA-680A, Toshiba Pty Ltd, North Ryde, NSW, Australia) with linear probe (PLT-704SBT, 7.5 MHz, 3.8 cm footprint) was used to scan isometric hip abductor activity in a custom-programmed application with strong contrast, high frame rate and highest sweep speed providing a sampling frequency of 450 Hz. The temporal resolution of the M-mode trace was 2.2 ms.

7.5.2.3.4 Data synchronization (in detail in section 3.4.6.5)

An electrical signal that was triggered by a footswitch was split to go synchronously into a separate channel of the Vicon data collecting system and into the ECG entrance of the US system to produce a visual signal on the bottom of the M-mode trace (Figure 7.15).

7.5.2.4 Data processing

7.5.2.4.1 Onset of torque and velocity of torque development

Torque onset was computer detected in a two-step procedure. First, a time window for onset detection was determined 300 ms before the signal exceeded the mean baseline torque amplitude plus fifteen standard deviations (SD). Within this window, torque onset was determined when the signal exceeded the mean baseline torque plus 5 SD. The velocity of torque development in the initial 200 ms starting from onset was determined by $F_{onset+200ms} - F_{onset} / 200$.

7.5.2.4.2 Onset of electrical excitation

EMG signals were processed using the Teager-Kaiser Energy Operator (TKEO) (Malone et al 2011, Solnik et al 2010, Li et al 2007b). The TKEO was applied to determine EMG onset in a comparable method as US onset, as a marked change of the energy level of the signal. In a time window starting 400 ms before torque onset, EMG onset was determined when the TKEO exceeded the mean of 400 ms baseline plus a threshold value. The threshold value depended on signal intensity, which varied between individuals. The appropriate threshold was determined per individual in three trials and set to 0.001, 0.01 or 0.05 V^2. Onsets were controlled visually.

7.5.2.4.3 Onset of M-mode high-frequency greyscale patterns

US M-mode frames including the synchronization signals and the visual occurrence of motion onset were cut from the recorded clips on the US system. Frames were labelled, saved, exported in tiff format and processed using a custom programmed

LabView application (V8.2.1., National Instruments, Texas, USA) (section 7.3). A central zone of one cm thickness within the deep gluteus medius and gluteus minimus was determined for analysis under consideration of a change of muscle thickness with activity. Depending on the total image depth, one cm was approximately 30 pixel-lines. Each pixel line was translated into greyscale amplitude and transformed using the TKEO. For each point in time, mean and standard deviation (SD) of the respective TKEO values were calculated (Figure 7.12). Three alternative M-mode onsets of high-frequency greyscale patterns were determined when the SD of the TKEO exceeded the amplitude of 800, 1200 and 2000 (see discussion of the unit in section 7.2.2.4) for 7 in 10 pixels.

7.5.2.4.4 Motion onsets preceding excitation onsets

In the biceps brachii muscle, Hug et al (2011b) demonstrated onset differences of up to 20 ms between adjacent channels of a multi-array surface electrode. These differences in excitation onset are an indication of the spatial heterogeneity of muscle recruitment. To account for local differences in muscle excitation, only motion onsets which preceded EMG onset for more than 10 ms were counted as being earlier than excitation.

7.5.2.5 Statistics

Data were evaluated for the subjects with and without hip symptoms separately. Blinded to the measurement results, an approximately gender and age matched, asymptomatic subsample was created for examining group differences. A reliability study was not included because onset determination was computed.

The relationship between EMG, M-mode and torque onsets of deep gluteus medius and minimus activity was examined by the latencies between EMG onset, the three M-mode US onsets of high-frequency greyscale patterns, SD800, SD1200 and SD2000, and torque onset. The latencies between EMG, M-mode and torque onset were averaged per subject, force level and velocity condition and described by mean, standard deviation (SD), standard error of measurement (SEM) and confidence intervals (CI). Onset agreement between EMG and M-mode was described for single trials by Bland-Altman plots and the limits of agreement (LOA).

Motion onsets preceding EMG onset were quantified for each of the three M-mode motion variables (SD800, SD1200 and SD2000). If trials with earlier motion onset could be explained by a preceding onset of the second hip abductor they were labelled

explained. Trials of early motion onset without preceding motion in the second hip abductor were labelled *unexplained*.

The comparison between the asymptomatic and the symptomatic subgroup could be only explorative. The small sample size (n = 4) did not justify statistical inferences. Results were presented for each subject by mean, SD and CI of the latencies between EMG and M-mode SD1200 onset relative to torque onset.

7.5.3 Results

7.5.3.1 Sample and data inclusion

Fourteen physiotherapists, seven female, were accepted for Study Five. Ten volunteers stated no hip symptoms. Four of the female volunteers reported occasional hip pain without specific diagnosis. One asymptomatic subject experienced marked pain during fine-wire insertion into gluteus medius that ceased after the insertion. Most trials of this subject indicated delayed and reduced gluteus medius motion (Figure 7.16). Data of this subject were excluded. Table 7.1 presents the sample characteristics.

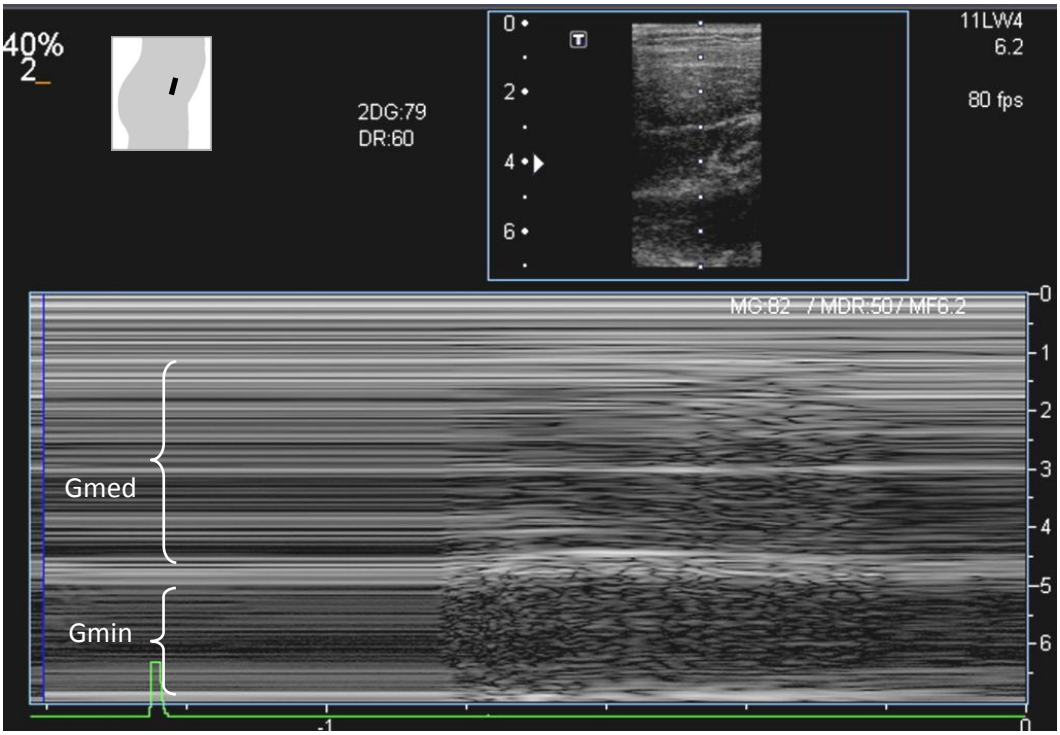


Figure 7.16. M-mode trace of isometric hip abductor activation (40% MVIC) of a subject who experienced marked pain during fine-wire electrode insertion into gluteus medius. Pain had ceased when the trial was performed. The M-mode trace indicates strongly delayed and reduced gluteus medius motion.

Table 7.1. Sample characteristics.

	Asymptomatic group	Symptomatic group	Asymptomatic subsample
gender	3 ♀, 7 ♂	4 ♀	3 ♀, 1 ♂
age	39.7 years	44.3 years	42.0 years
BMI	23.2 kg/m²	24.0 kg/m²	21.5 kg/m²

Altogether, 143 trials were recorded. The rate of detection of the M-mode high-frequency variables differed with highest occurrence for the SD800 motion onset and lowest for the SD2000 motion onset and with higher occurrence in gluteus minimus than in gluteus medius (Table 7.2). With higher force level and velocity, the detection rate increased. The low rate of EMG onset detection in the symptomatic subjects was apparently caused by a reduced ability to relax the gluteus medius muscle. Trial inclusion was when EMG and M-mode onsets could be detected (Table 7.3).

Table 7.2. Number and percentage of trials in which EMG onset or M-mode high-frequency onset variables SD800, SD1200 and SD2000 could be detected.

	Asymptomatic group (n = 9)				Symptomatic group (n = 4)			
	EMG	SD800	SD1200	SD2000	EMG	SD800	SD1200	SD2000
Deep Gmed	86 (87%)	82 (83%)	72 (73%)	40 (40%)	25 (57%)	32 (73%)	28 (64%)	25 (57%)
Gmin	96 (97%)	86 (87%)	73 (74%)	46 (46%)	41 (93%)	38 (86%)	31 (70%)	15 (34%)

Table 7.3. Trial inclusion.

	Asymptomatic subjects			Symptomatic group			Asymptomatic subsample		
	SD800	SD1200	SD2000	SD800	SD1200	SD2000	SD800	SD1200	SD2000
Deep Gmed	70 (71%)	62 (63%)	36 (36%)	23 (52%)	22 (50%)	21 (48%)	32 (73%)	31 (70%)	22 (50%)
Gmin	84 (85%)	72 (73%)	45 (45%)	37 (84%)	30 (68%)	14 (32%)	43 (98%)	39 (89%)	29 (66%)

7.5.3.2 Relationship between EMG, M-mode high-frequency patterns and torque onset

The latencies between the onsets of EMG, M-mode high-frequency greyscale patterns and torque are presented in Table 7.4 (full results including standard deviations in Appendix 7.1). Latency variation was higher relative to torque onset than to EMG (Table 7.5). Latencies were more consistent in the 60% MVIC trials with fast

activation velocity. The LOA for EMG and high-frequency pattern onsets were wide (Figure 7. 17 a-d and Table 7.6). M-mode high-frequency patterns were indicative of gluteus medius excitation in more than 90% (SD1200) or in 100% (SD2000) and of gluteus minimus excitation in 75% (SD1200) or in 86% (SD2000) (Table 7.7). In spite of most determination coefficients > 0.8, the precision of predicting excitation onset from high-frequency pattern onset was low (Table 7.6). Figure 7.18 visualizes onset relationships in an asymptomatic subject.

Table 7.4. Latencies of M-mode high-frequency pattern and torque onset relative to EMG onset, averaged per force level and velocity condition; +latency, onset followed EMG onset.

Mean (±CI), SEM, ms	Asymptomatic group, n = 9			
	SD800	SD1200	SD2000	F
dGmed to EMG	26 (±17) 34	51 (±25) 49	83 (±66) 105	72 (±38) 76
Gmin to EMG	0 (±22) 45	51 (±38) 78	62 (±43) 74	102 (±31) 64
dGmed fast to EMG	10 (±11) 11	16 (±13) 13	103 (±153) 156	48 (±18) 19
Gmin fast to EMG	11 (±18) 20	75 (±94) 102	38 (±71) 72	53 (±26) 28

Abbreviations in tables 7.4 – 7.8: CI, confidence intervals; dGmed, deep gluteus medius; EMG, electromyography; F, torque onset; SEM, standard error of measurement.

Table 7.5. Latencies of M-mode high-frequency pattern onset relative to torque onset, averaged per force level and velocity condition; +latency, onset followed torque onset; -latency, onset preceded torque onset.

Mean (±CI) SEM, ms	Asymptomatic group, n = 9		
	SD800	SD1200	SD2000
dGmed to torque	-54 (±27) 89	-28 (±31) 93	15 (±63) 144
Gmin to torque	-96 (±29) 97	-35 (±36) 11	-17 (±19) 46
dGmed fast to torque	-41 (±16) 17	-32 (±13) 14	59 (±124) 135
Gmin fast to torque	-43 (±25) 27	21 (±80) 87	-14 (±18) 18

Table 7.6. Prediction of excitation onset from high-frequency pattern onset: mean difference, limits of agreement (LOA) and determination coefficient (r^2); single trials.

Mean difference, LOA in ms, r^2	Asymptomatic group, n = 9		
	SD800	SD1200	SD2000
dGmed to EMG	21, 90, *0.98*	46, 140, *0.95*	91, 349, *0.33*
Gmin to EMG	0, 184, *0.90*	53, 267, *0.80*	43, 165, *0.94*

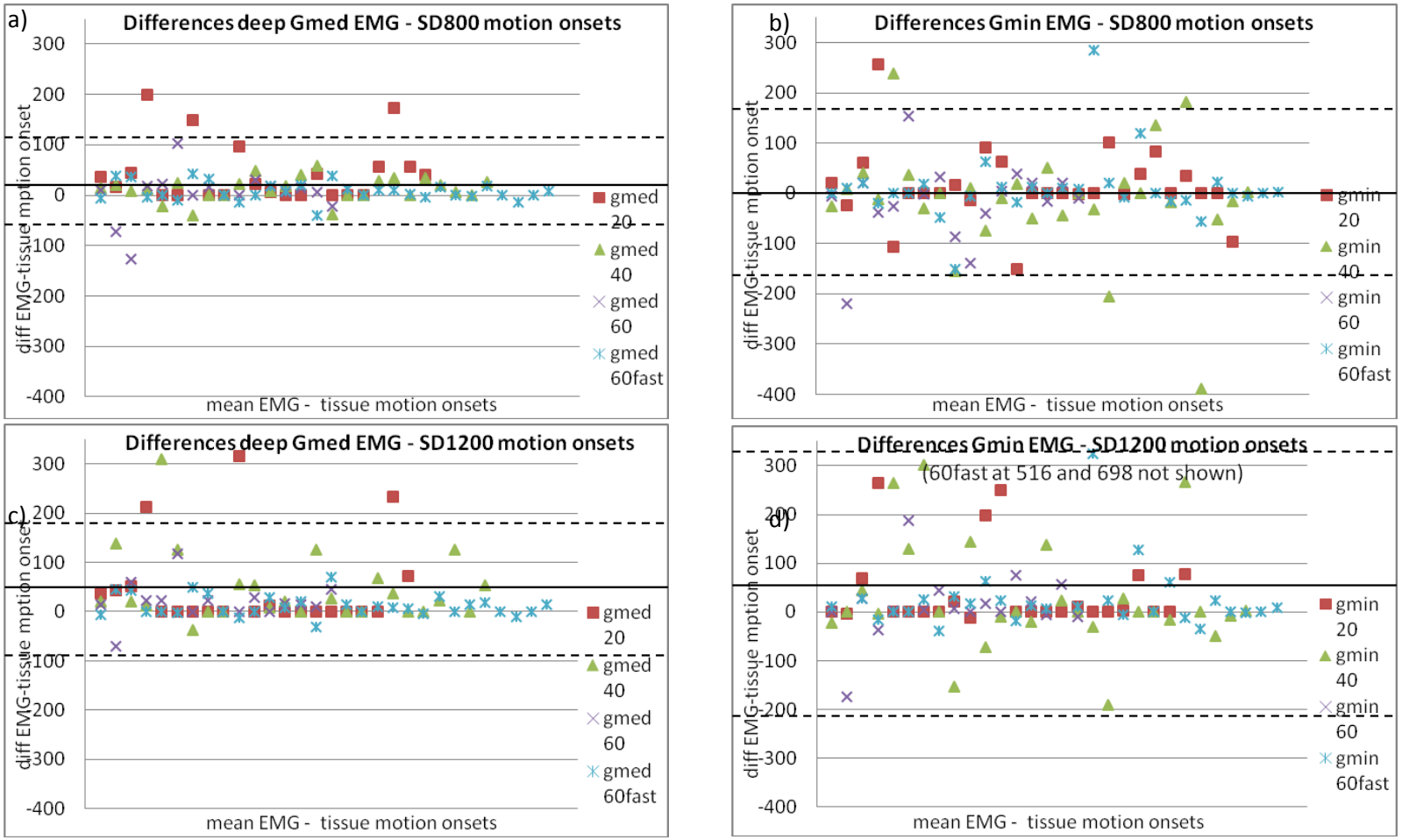


Figure 7.17 a-d. Bland Altman plots of the differences between EMG and M-mode rapid motion onsets, SD800 and SD1200 motion variables, single trials according to force level and velocity condition, LOA dashed lines. Data of the asymptomatic subjects, n = 9.

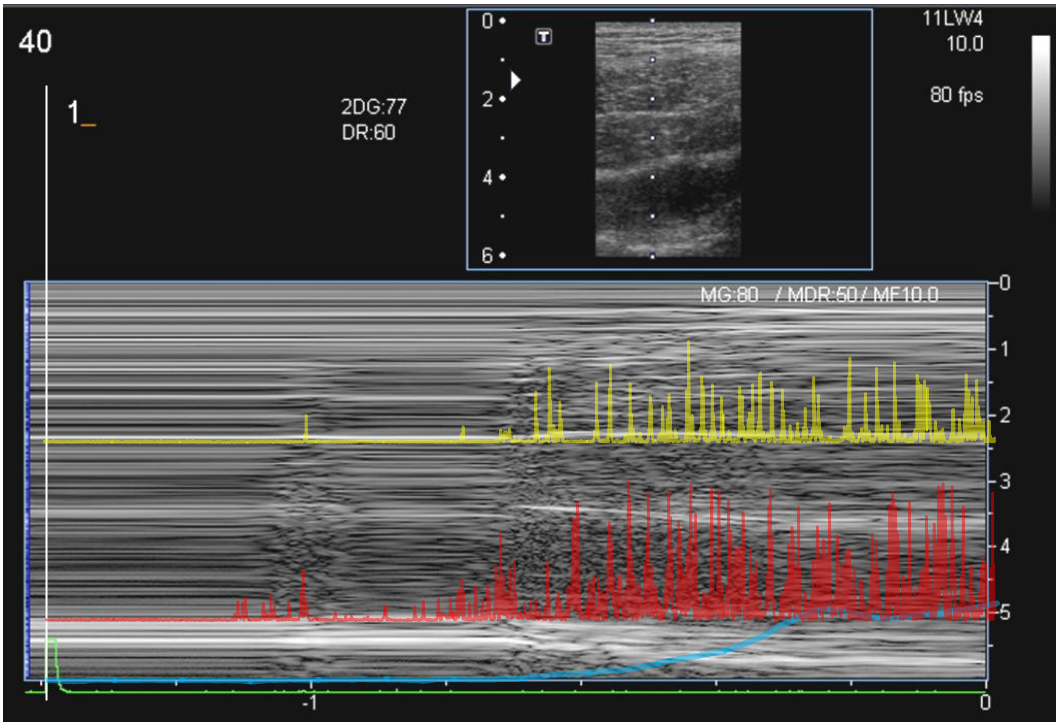


Figure 7.18. Synchronized signal traces of an asymptomatic subject, M-mode trace of Figure 7.15; blue, torque¸ red, fine-wire EMG of gluteus minimus; yellow, fine-wire EMG of deep gluteus medius.

7.5.3.3 Motion onsets preceding EMG onset

Table 7.7 presents the percentage of trials in which high-frequency pattern onset preceded EMG onset. Unexpectedly, most of these early motion onsets could not be explained by earlier motion in the second measured muscle, nor by spatial heterogeneity of muscle recruitment in the range documented by Hug et al (2011b).

Table 7.7. Percentage of trials in which M-mode high-frequency onset preceded EMG onset by more than 10 ms, in brackets percentage of trials in which earlier motion may be explained by motion in the adjacent muscle, *in italics* mean time of advance of the unexplained trials.

	Asymptomatic group (n = 9)		
	SD800	SD1200	SD2000
dGmed	12.9% (33.3%) *43 ms*	8.1% (60%) *12 ms*	0.0% (n.a.)
Gmin	38.1% (21.9%) *63 ms*	22.2% (25%) *63 ms*	15.6% (14.3%) *38 ms*

7.5.3.4 Differences between the symptomatic and the matched asymptomatic subjects

The comparison between the four symptomatic subjects and the matched subsample indicated tendencies for differences in the relationship of excitation and muscle motion onset. While the asymptomatic subjects indicated EMG onsets preceding high-frequency pattern onsets, the symptomatic subjects demonstrated a less consistent relationship between EMG and high-frequency pattern onset (Figure 7.19 and Table 11.5 in Appendix 7.1) and larger variability of EMG onset relative to torque onset. In the symptomatic subsample, onset of M-mode high-frequency pattern preceded EMG onset in a higher percentage of trials (Table 7.8).

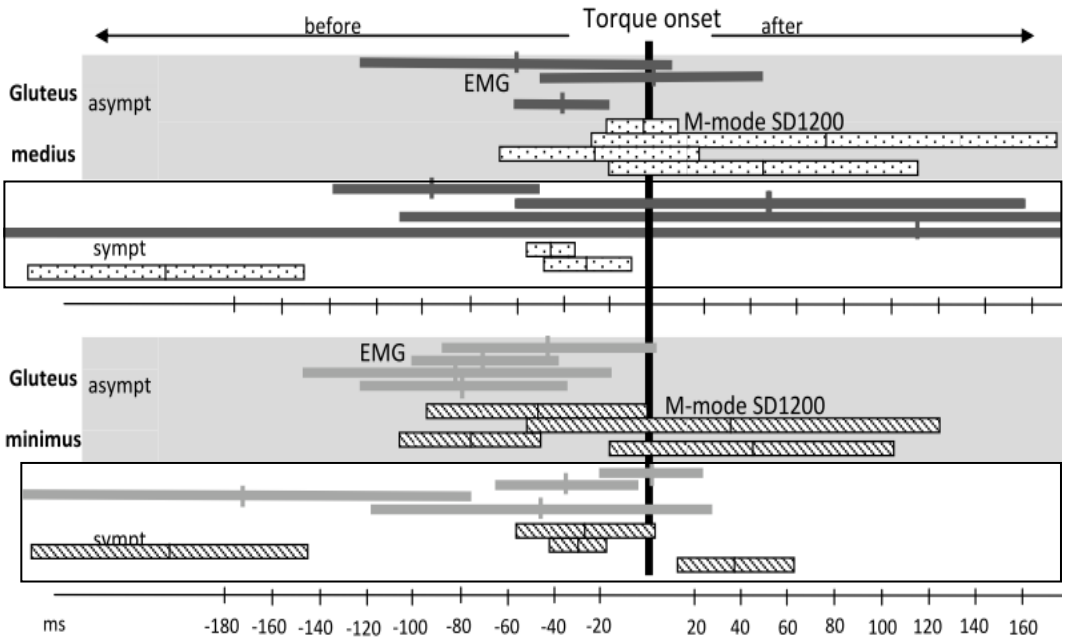


Figure 7.19. Subsample comparison, mean ± CI of EMG onsets (grey bars) and M-mode high-frequency pattern onsets (variable SD1200, dotted and shaded bars) relative to torque onset (bold vertical line) demonstrate a less consistent relationship in symptomatic subjects. Data are documented in Table 11.5 in Appendix 7.1.

Table 7.8. Subsample comparison, the percentage of trials in which rapid tissue motion preceded EMG onset by more than 10 ms was larger in symptomatic subjects. In brackets percentage of trials in which earlier motion could be explained by motion in the adjacent muscle and mean time of advance of the unexplained trials.

	Symptomatic group			Asymptomatic subsample		
	SD800	SD1200	SD2000	SD800	SD1200	SD2000
Gmed	47.8% (54.5%, 116 ms)	40.9% (77.8%, 129 ms)	42.9% (44.4%, 159 ms)	18.8% (16.7%, 44 ms)	6.5% (0.0%, 12 ms)	0.0% (n.a.)
Gmin	54.1% (30.0%, 76 ms)	36.7% (45.5%, 60 ms)	7.1% (100%)	46.5% (15.0%, 45 ms)	23.1% (11.1%, 41ms)	13.8% (0.0%, 13 ms)

7.5.3.5 Do gluteus minimus and gluteus medius activate in sequence?

In the asymptomatic subjects, the fine-wire EMG measurements in Study Five confirmed preceding activation of the deepest hip abductor gluteus minimus (suggested by M-mode US measurements, Chapter Six, section 6.1.3.3). Onset of gluteus minimus excitation started in mean 34 (SD 118, CI ±25) ms before gluteus medius, *p* .011. The M-mode study presented in Chapter Six found a difference of 36 ms.

7.5.4 Discussion

This study explored an innovative method to assess the timing of deep muscle activity non-invasively by the onset of high-frequency greyscale patterns in M-mode US traces. High-frequency greyscale patterns indicate rapid muscle motion, but the measurement is also influenced by the diameter of muscle fascicles. The hypothesis was that M-mode high-frequency patterns reflect electrical excitation of muscle.

7.5.4.1 High-frequency variable selection

Three variables that were defined by different intensities of high-frequency patterns were tested and differed in applicability. The SD800 variable was detected in most trials and was the best predictor of EMG onset. However, in gluteus minimus in more than 38% of trials the SD800 variable was not associated with concurrent electrical excitation. The M-mode high-frequency variables SD 1200 and SD200 were superior for detection of electrical excitation (Table 7.7). The SD2000 variable was detected only in approximately half of trials, an indication of minor relevance. None of the M-mode high-frequency variables enabled a precise prediction of EMG onset in both muscles and under all measured conditions. The presentation of the results focuses on the SD1200 variable, which was detected in approximately 70% of trials, was in 78% (gluteus minimus) to 92% (gluteus medius) indicative of electrical muscle excitation and most indicative of subgroup differences.

7.5.4.2 Relationship between EMG and M-mode high-frequency pattern onset

Study Five confirmed that the onsets of EMG and M-mode high-frequency patterns are highly correlated. Studies Four and Five both indicate that the latency between electrical excitation and muscle motion is not a fixed time span but variable. The variation of latencies was supported by the explorative study design which included

multiple force levels and two velocities of activation. In the small sample of an invasive fine-wire EMG study, high variation is a statistical disadvantage that precluded the investigation of differences between force levels in Study Five. The advantage of the study design was that it informed about limitations in the application of M-mode high-frequency variables. In conclusion, in a study task with self selected velocity of performance and lower force levels, M-mode high-frequency patterns cannot be used to infer electrical muscle excitation. In general, in higher force levels and rapid task performance M-mode high-frequency patterns are good indicators of concurrent electrical excitation (Tables 7.4 and 7.5, Figure 7.20).

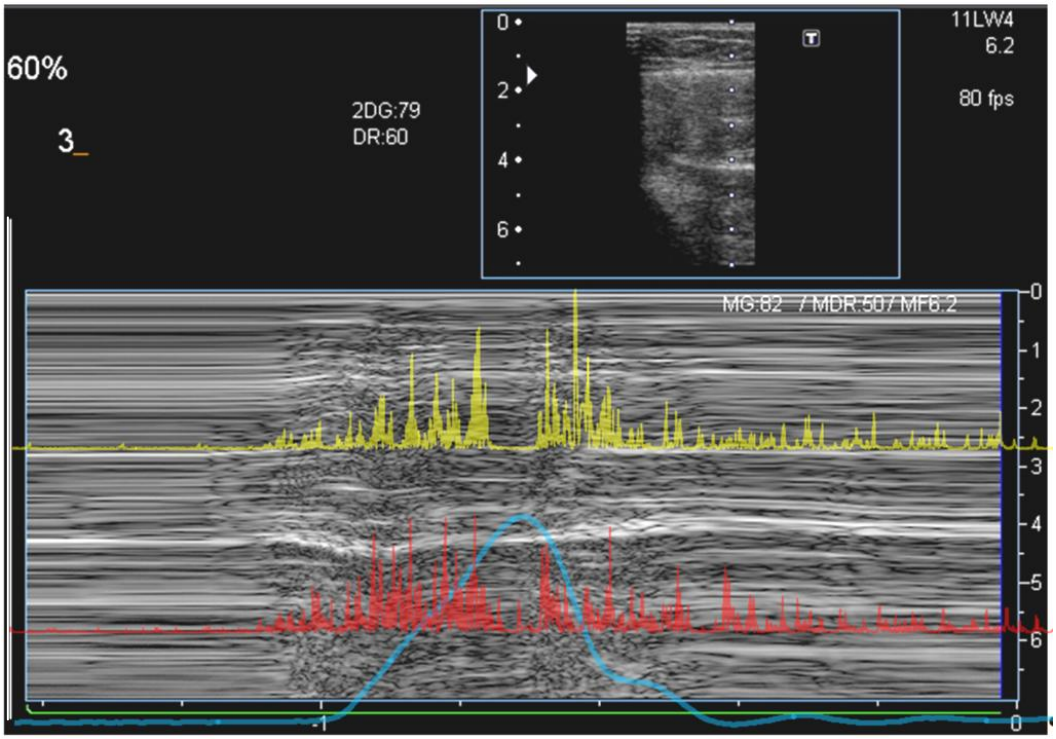


Figure 7.20. The M-mode trace of a rapid, short 60% MVIC trial (blue, torque) indicates the close relationship between EMG measured activity (red, gluteus minimus fine-wire EMG, yellow, deep gluteus medius fine-wire EMG) and M-mode registered muscle motion.

7.5.4.3 Muscle motion onsets preceding EMG onset

In self-selected abduction velocity, all but one subject produced trials in which the onset of high-frequency patterns, i.e. rapid muscle motion preceded EMG onset. How can rapid muscle motion occur before EMG onset? It has been suggested that the transmission of motion of an adjacent muscle, here the gluteus medius muscle, caused motion in the measured muscle (Mannion et al 2008a, Vasseljen et al 2006).

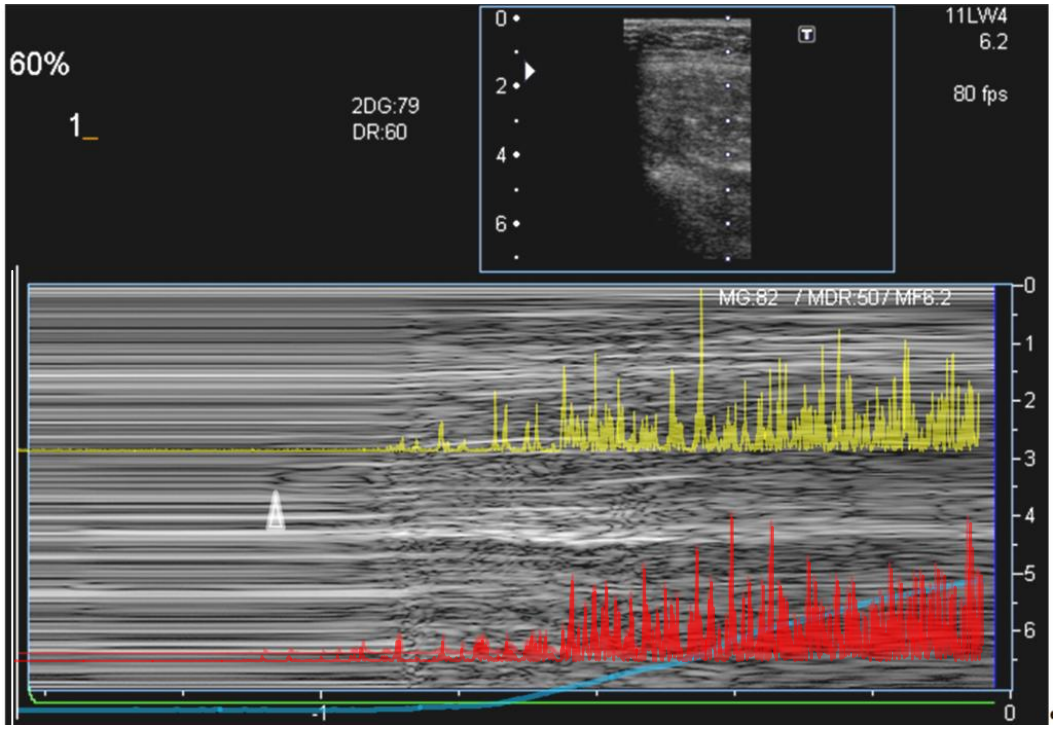


Figure 7.21. Earlier gluteus medius motion (arrowhead) could not be explained by earlier motion of adjacent muscles in the scanned tissue section. In this trial, earlier motion appears to result from an isolated gluteus medius twitch that was not reflected in motor unit activity in the fine-wire sampling area.

In Study Five, the smaller part of early motion onsets was explainable by motion transmission from gluteus medius (Tables 7.7 and 7.9). In twelve trials, high-frequency pattern in gluteus minimus started before electrical excitation, in spite of later gluteus medius motion. It appears unlikely that the measured motion in gluteus minimus was caused by vessel pulsation, which has not been observed to produce high-frequency patterns (p. 135, Figure 6.7). It is not probable that the early motion onsets are transducer motion artefacts, which should be visible and measurable over the full depth of measured muscles (pp. 135 – 137, Figures 6.8 – 6.11). Three sources of gluteus minimus motion appear possible, beside excitation at the measured location or in gluteus medius.

(1) Pelvic motion. The study was set up deliberately to avoid pelvic motion. The pelvis was fixed on the contralateral side and the contralateral leg was rested in flexion on a foam block to reduce possible bracing mechanisms. The isometric abduction task required only minimal limb motion. Nevertheless, pelvic motion cannot be excluded.
(2) Gluteus minimus motion in an adjacent muscle part out of the scanning plane.

(3) Excitation of adjacent motor units which were not represented in the small sampling area of the fine-wire electrodes (Figure 7.21). Hug et al (2011b) reported timing differences of up to 20 ms between adjacent regions of the biceps muscle, Brown et al (2007) identified a difference of up to 31 ms between adjacent deltoid segments. The spatial heterogeneity of gluteus medius and minimus excitation is unclear.

The phenomenon of muscle motion before EMG onset requires further research.

7.5.4.4 Subgroup differences

Unexpectedly, this first exploration of M-mode high-frequency patterns and electrical muscle excitation demonstrated differences in the relationship between EMG and muscle motion in the four subjects with hip symptoms. Due to the small number of symptomatic subjects, a different relationship between electrical excitation and muscle motion cannot be generalized. Nevertheless, this preliminary finding is of great importance and provides a justification for an invasive study on symptomatic subjects. Existing studies in which EMG and US measurements of muscle activity have been compared were performed on asymptomatic subjects (Vasseljen et al 2009, Mannion et al 2008a, Pulkovski et al 2008, Rudroff et al 2008, Shi et al 2008, Kiesel et al 2007b, Vasseljen et al 2006, McMeeken et al 2004, Hodges et al 2003). To the author's knowledge, no comparison between EMG and US measurements on symptomatic subjects has been published (July 2012). The current stage of knowledge does not allow for comparing muscle activity patterns of US studies with those of EMG studies for subjects with pathology.

7.5.4.5 Limitations

Latency variation in Study Five was larger than in literature reports (Vasseljen et al 2009, Mannion et al 2008a, Pulkovski et al 2008, Vasseljen et al 2006), likely caused by the range of included force levels and self-selected activation speed. Variation may have been also increased due to computed onset detection (Uliam Kuriki et al 2011). Latency variation of the asymptomatic subjects was well in the range reported in the literature for the 60% trials in rapid abduction. Some disagreement of measurements was potentially caused by scanning a slightly different muscle part by US than was measured by EMG. Once the fine-wire electrodes are inserted, it is difficult to scan their exact position. It is unlikely that these slight differences in transducer position led to the observation of significant group differences.

The most important limitation of this study is the small sample size of the group comparison. For ethical reasons and due to practical difficulties to find sufficient subjects, fine-wire EMG studies consist of small samples. The results of the subgroup comparison have preliminary character. Nevertheless, the results have two important implications. First, the relationship between EMG and US measurements of activity may differ between asymptomatic and symptomatic subjects. These differences in relationship need further investigation. Second, non-invasive M-mode high-frequency patterns may indicate differences of clinical relevance in activity related muscle motion between subjects with and without hip pathology.

7.5.5 Conclusions

M-mode US traces of activity related muscle motion reflect a mechanical component of the process of muscle activation which complements the information provided by EMG which reflects the electrical component of muscle activation. The relationship between EMG and M-mode high-frequency patterns is close, variable and may be affected by pain. Further research is necessary to explore the clinical use of the information provided by M-mode US on the mechanical aspect of muscle activity.

7.5.6 Acknowledgements

Special acknowledgement is due to Paul Davey who confirmed the author's decision to use the TKEO algorithm to analyse the US data/scans from a physicist's view and who programmed the LabView based image analysis. The contribution of Dr Will Gibson, who inserted the fine-wire electrodes, is gratefully acknowledged. Further acknowledgement is extended to Toshiba Medical Systems, Australia, for the loan of the physiology unit which enabled the synchronization of EMG and US data.

7.6 Ultrasound measurements of onset of muscle activity cannot replace electromyographic measurements. Are they worth further research?

The studies in Chapters Six and Seven suggest that the physiological process of excitation-contraction coupling is not the only source of motion in activating muscle groups. Some mechanical phenomena could not be explained by local excitation. The translation of excitation into mechanical activity appears to be affected by other, yet undetermined factors. The current concept of muscle activity that is based on electrical measurements is limited for the interpretation of mechanical phenomena.

At this point, one may discard US recordings of muscle activity, as they did not indicate a clear and unambiguous sign of the level or the timing of the electrical excitation of muscle. To leave US imaging of muscle activity means to stay with the assessment of deep muscle activity by fine-wire EMG. The following section reports an evaluation of the experience of the fine-wire EMG study.

7.6.1 Evaluation of the experience of the fine-wire EMG data collection

Directly after Study Five, all subjects received a questionnaire which was designed to explore the experience and potential side effects of fine-wire insertion and to improve the performance of a fine-wire study (Appendix 7.2). Eleven of fourteen questionnaires (79%) were returned.

No subject experienced a sympathetic reaction on needle insertion. Five subjects stated muscle soreness after insertion. Duration of soreness was described between a couple of hours to 3-4 days. One subject reported bruising and one minor swelling of the insertion spot after the procedure. Table 7.9 indicates the pain scores allocated to four distinct phases of the fine-wire insertion process (pain scored between 0 = no pain and 10 = maximal pain).

Table 7.9. Pain scores reported by the subjects who participated in the fine-wire EMG setup.

	Skin penetration	Placing the needle into the muscle	Deep fascia penetration	Needle withdrawal
Mean pain (SD)	1.7 (1.0)	2.7 (2.7)	4.6 (2.0)	2.1 (1.5)

From the perspective of the study subjects, the penetration of the gluteus minimus aponeurosis was the most critical event during fine-wire electrode insertion into the gluteus medius and minimus muscles.

From the perspective of the investigators, the insertion of fine-wire electrodes into the hip abductors included muscle-specific difficulties. One difficulty was the sensitivity of the gluteus minimus aponeurosis. A second difficulty was pain with weight-bearing. Measurements in single leg stance had been planned (Chapter Three, section 3.2) and were not feasible. Due to this limitation, the original study design, which included a step down task, had to be changed.

The removed electrodes demonstrated deformations (Figure 7.22), an indication of strong forces within the activating tissue. In Chapter Five, section 5.1.4.5 transverse motion of the gluteus medius more than minimus has been reported. Possibly, the transverse motion of one muscle against the other was a source of mechanical irritation which limited the subjects' active performance. Under these conditions the possibility that fine-wire measured muscle activation may be influenced by nociception is increased.

Fine-wire EMG is limited for the assessment of deep hip muscle activation. Compared to EMG, US imaging of muscle activity is a new and minimally explored technique which may, with further research, develop to be a useful tool to understand the mechanical side of muscle activity. The further exploration and investigation of US measurements of hip muscle activity are necessary to gain information about the detailed functioning and coordination of the deep hip muscles.

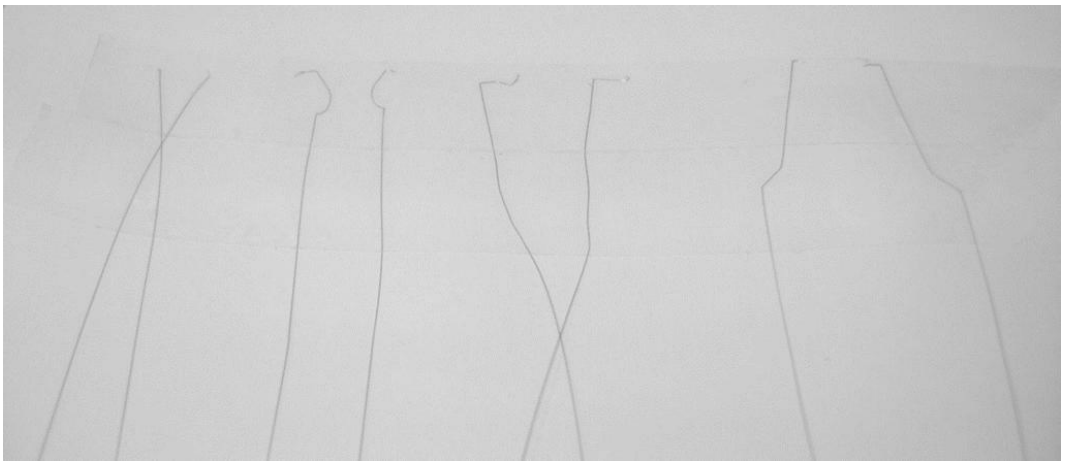
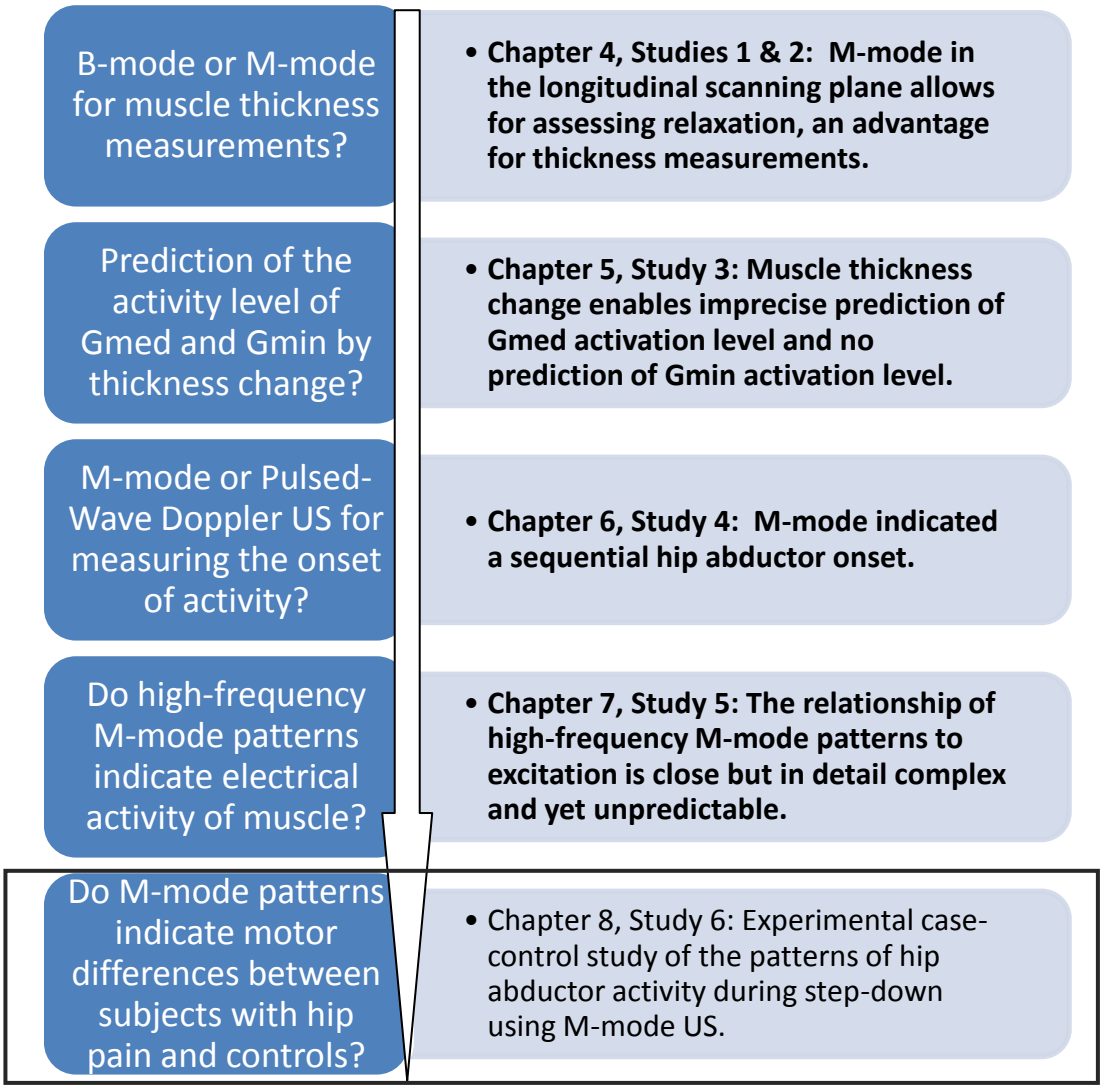


Figure 7.22. Deformation of removed fine-wire electrodes, left electrode pair is before insertion.

7.7 Summary of the research findings of Chapters Six and Seven on onset measurements by M-mode US

M-mode US enables the identification of the location of activity related muscle motion for the full depth of the image. M-mode traces of gluteus medius and minimus activity during isometric hip abduction demonstrated asynchronicities of the onsets of muscle motion with high temporal resolution and non-invasively. Differential motion onsets support functional differences between the gluteus minimus muscle, the deep gluteus medius and the superficial gluteus medius muscle. The M-mode traces of muscle motion were clearly related to muscle excitation and to torque output, although the precise relationship between the electrical and the mechanical signals of hip abductor activity was not predictable.

Study Five introduced innovative M-mode US motion variables characterised by high-frequency greyscale patterns. The study findings suggest that M-mode high-frequency patterns indicate differences in the activity patterns of superficial and also deep hip muscles between subjects with hip pain and controls.


B-mode or M-mode for muscle thickness measurements?
• Chapter 4, Studies 1 & 2: M-mode in the longitudinal scanning plane allows for assessing relaxation, an advantage for thickness measurements.
Prediction of the activity level of Gmed and Gmin by thickness change?
• Chapter 5, Study 3: Muscle thickness change enables imprecise prediction of Gmed activation level and no prediction of Gmin activation level.
M-mode or Pulsed-Wave Doppler US for measuring the onset of activity?
• Chapter 6, Study 4: M-mode indicated a sequential hip abductor onset.
Do high-frequency M-mode patterns indicate electrical activity of muscle?
• Chapter 7, Study 5: The relationship of high-frequency M-mode patterns to excitation is close but in detail complex and yet unpredictable.
Do M-mode patterns indicate motor differences between subjects with hip pain and controls?
• Chapter 8, Study 6: Experimental case-control study of the patterns of hip abductor activity during step-down using M-mode US.

8 Chapter Eight: M-Mode High-Frequency Greyscale Patterns Indicate Altered Hip Abductor Activity in Subjects with Hip Pain

Overview of Chapter Eight

- Section 8.1 introduces the topic of anterior, non-traumatic hip pain and presents the rationale behind Study Six.
- Section 8.2 describes the methods of Study Six in which motor patterns of hip abductor activity of subjects with and without hip pain were compared using M-mode US high-frequency variables.
- Section 8.3 presents the results of Study Six.
- Section 8.4 discusses the results, their limitations and the implications of Study Six for the physiotherapy treatment of hip pain and for the further development of M-mode US measures of muscle activity.

Study Six: Do M-mode high-frequency greyscale patterns indicate altered hip abductor activity in subjects with hip pain?

8.1 Introduction

8.1.1 Relevance of anterior hip pain

The prevalence of hip pain in older Caucasian populations has been reported between 11.9% who had pain in the past four weeks (Cecchi et al 2008) and 19.2% who had pain in the past year (Dawson et al 2004). Hip pain is not confined to the older age group. Occasional or permanent hip pain was reported by 6.5% of an adolescent population sample (Spahn et al 2005).

Hip pain is a symptom associated with a range of diagnoses and it may arise from the hip joint itself, from periarticular structures, from the lumbar spine, the sacroiliac joint, the symphysis pubis or other sources (Bierma-Zeinstra et al 2001). Recent developments in the diagnosis of hip pain include the description of the 'femoroacetabular impingement' (FAI) (Ganz et al 2003, Ito et al 2001, Klaue et al 1991) and the increased use of magnetic resonance imaging (MRI) and hip arthroscopy to identify soft tissue lesions, in particular of the acetabular labrum (Tibor and Sekiya 2008, Blankenbaker and Tuite 2006, Mengiardi et al 2006, Troum and Crues 2004, Mitchell et al 2003). Figure 8.1 represents a typical distribution of hip symptoms and related diagnoses (Segal et al 2007, Birrell et al 2005, Lievense et al 2005, Birrell et al 2003, Bierma-Zeinstra et al 2001).

Anterior hip pain in the groin is specifically relevant with regard to hip joint degeneration. In the Caucasian ethnicity, osteoarthritis (OA) of the hip joint affects between 8 and 15% of the population aged over 55 (Dagenais et al 2009). Joint degeneration and the associated limitations in musculoskeletal performance and social participation are a health burden of increasing importance in the ageing Western populations (Buckwalter et al 2004).

Figure 8.1. A typical distribution of hip symptoms and diagnoses in a sample of 100 subjects with hip pain, based on a cluster analysis of 224 consecutive patients with hip pain (Bierma-Zeinstra et al 2001), complemented by the referenced literature. The figure includes considerations for subject recruitment for Study Six.

Current medical treatment focuses on hip joint replacement, a common treatment which should be the last option. In Germany, an increase of 46% in first implantations of hip prostheses was registered in 2008 (BQS 2009). A volume of 210,000 hip prostheses were implanted in 2009, 20% in patients under 60 years (Schnabel and Borelli 2011). The direct costs for first implantations were estimated between 12 and 13 million Euro, not including costs for complications and follow-up (Röttger 2012). The impact and the costs of hip joint degeneration need to be reduced by preventive strategies and by a targeted, conservative treatment regime to preserve mobility and delay joint replacement. The identification of individuals at risk or in early stages of joint degeneration is necessary.

FAI, labral pathology and subtle, femoral deformities up to hip dysplasia are considered functionally linked predecessors of hip joint degeneration (Lequesne and Bellaïche 2012, Freehill and Safran 2011, Harris-Hayes and Royer 2011, Ganz et al 2003, Wagner et al 2003). Pain in the groin area during loading is with high likelihood associated with the hip joint (Plante et al 2011, Holmich and Dienst 2006, Brown et al 2004). The groin is the prevalent pain location in femoroacetabular impingement (Ejnisman et al 2011), labral pathology (Burnett et al 2006) and the typical pain location in hip osteoarthritis (Bierma-Zeinstra et al 2001). Recurrent, non-traumatic, anterior hip pain in middle or older age can be considered to indicate an increased risk to develop hip joint degeneration or may be a sign of the disease (Cibulka et al 2009).

8.1.2 Pathogenesis

In FAI and labral pathology, the central role of active hip joint mechanics is well recognized (Leunig and Ganz 2009). As stated by Martin and Tashman (2010, p. 252) “FAI is, by nature, a dynamic disorder whereby soft-tissue damage results from abnormal motion of the femur relative to the acetabulum.” The abnormal motion in FAI is the forceful contact of the femoral neck against the acetabular labrum in flexion or internal rotation / adduction movements. The acetabular labrum is increasingly damaged over the course of the pathology, and the acetabular ‘bump’ leads to bony appositions on the femoral neck that further reduce the range of movement (Lequesne and Bellaïche 2012, Martin and Tashman 2010, Nepple et al 2009). The pain provocative

and destructive mechanics of FAI are well described, but not predisposing, functional deficits that may cause abnormal motion.

Most literature reflects the understanding that FAI is caused by a combination of bony abnormalities and the repetitive performance of end grade hip movements (Lequesne and Bellaïche 2012, Kennedy et al 2009). Current therapy focuses on surgical or arthroscopic corrections of the associated structural deformations (Lequesne and Bellaïche 2012, Huang and Parvizi 2011, Matsuda et al 2011). Limitations of this *structural approach* have been demonstrated by studies that report a poor prediction of hip pain or OA from the alpha angle and the head-neck offset, structural hip joint parameters that are considered indicative of FAI (Audenaert et al 2011, Gosvig et al 2008, Van der Heijden et al 2008).

A *functional approach* needs to consider the control of femoral motion and local hip joint loading. Evidence of reduced hip joint loading by muscle activity has been provided by Bergmann et al (2004) who found that anticipation of stumbling reduced the measured contact forces within the hip joint to less than half. Significantly lower peak hip moments during walking with increased ankle push off were reported by Lewis and Ferris (2008). General hip muscle weakness and conclusively a reduced ability to reduce hip joint loading actively has been identified in subjects with FAI (Casartelli et al 2011). Clinical experience suggests that the functional deficits associated with FAI are more specific than a general lack of strength.

Sahrmann (2002) proposed that not only contact forces but also accessory motions of the femoral head have to be controlled in hip joint impairments. The hip joint is incongruent (von Eisenhart-Rothe et al 1999, Menschik 1997, Neusel et al 1996), a configuration which provides space for active joint play and accessory motions. Investigations in eight cadavers indicated a physiological anterior-posterior glide of the femoral head in the acetabulum (Harding et al 2003). An MRI based study on thirty professional dancers demonstrated a regular translation of the hip joint centre in maximal hip flexion (Gilles et al 2009). Research on other joints supports the role of muscular control of accessory joint motion. Altered arthokinematics and insufficient control of the translation of the centre of rotation is assumed to be one determinant of glenohumeral impingement (Michener et al 2003). The control of secondary motions in the knee joint has been identified as a preconditioning factor in the development of

cartilage degeneration and subsequent OA of the knee joint (Andriacchi and Mündermann 2006, Andriacchi et al 2004)Joint play and a well-defined mixture of rolling and gliding motions at mechanically suited joint loci have been described as the physiological condition of joint lubrication (Kubein-Meesenburg et al 1993, Kubein-Meesenburg et al 1990). It can be assumed that excessive as well as reduced joint play and an alteration in the rolling and gliding motion components have long-term, negative effects on joint integrity. A well-defined mixture of rolling and gliding motions can be translated into an appropriate control of the centre of rotation or in dynamic joint stabilization. Abnormal femoral motion and inappropriate joint loading in FAI may result from impaired control of the centre of rotation. It is essential to understand which structures and which muscle activities ascertain the control of the hip joint centre in pain provocative movements.

8.1.3 A motor control approach

A minor part of hip pain related literature reflects a motor control approach (Wright and Hegedus 2012, Yazbek et al 2011, Austin et al 2008, Zimny 1998). These authors emphasized the control of dynamic valgus, a combined femoral adduction and internal rotation movement. The control of dynamic valgus is mainly achieved by activity of the abductors and external rotators, many of which are deep lying. Their activity is not assessable by surface EMG. Deep lying muscles are thought to have a stronger importance in joint stabilization than superficial muscles which are thought to be prime movers (Hamilton and Richardson 2001). The abductors gluteus medius and minimus control frontal plane pelvic and hip motion. The smaller lever length of gluteus minimus (Dostal et al 1986), the shorter length of gluteus minimus fibres (Friederich and Brand 1990) and its greater content of muscle spindles (Stillman 2000) and slow twitch fibres (Hitomi et al 2005) support the assumption that the prime function of gluteus minimus is the exact positioning and stabilization of the acetabulum on the femoral head, rather than leg abduction.

Currently, the only direct evidence of a differential recruitment of gluteus minimus relative to gluteus medius is from an MRI study of the metabolic response to isometric abduction and one-legged stance (Kumagai et al 1997). Based on muscle thickness measurements by MRI and on surface EMG studies, altered patterns of hip

abductor activation with hip OA have been found between the superficial gluteus medius and tensor fasciae latae muscles (Grimaldi et al 2009b, Sims et al 2002, Long et al 1993) and also between gluteus medius and minimus, and even between different parts of the gluteus medius muscle (Grimaldi et al 2009a, Grimaldi et al 2009b).

These literature reports refer to a redistribution of force sharing between the hip abductors. It is not known, but likely, that the timing of hip abductor activation is also altered between synergists in painful conditions (Hodges and Tucker 2011). Study Five in Chapter Seven suggested that M-mode high-frequency patterns enable clinically relevant measurements of the onset of gluteus medius and minimus activity. The following study addresses the need to assess the patterns of abductor muscle activity in subjects with anterior hip pain and controls non-invasively. The objective of Study Six was to determine, by M-mode US onsets of high-frequency patterns, whether the mechanical activity of the superficial and deep gluteus medius and gluteus minimus differs between subjects with non-traumatic anterior hip pain and controls.

8.2 Methods

8.2.1 Participants

The Curtin University Human Research Ethics Committee approved this study (PT0150/2009). Sample size calculations based on the results of Study Five indicated that a sample of 34 subjects per group would provide 80% power to detect a difference of 62 ms between groups when the SD is 90 ms.

Subjects and controls were recruited by radio advertising in a broadcast station that focussed on the 50+ age group, by posters in radiography and physiotherapy clinics and in the University (Appendices 8.1 and 8.2). Interested individuals between 35 and 65 years of age were screened by telephone for inclusion (Appendices 8.3 and 8.4).

Inclusion criteria for subjects with pain were recurrent hip pain with insidious pain onset within the area described by Birrel et al (2005) and predominantly in the groin. Inclusion criteria for controls were no hip or knee pain. As with the previous studies, the exclusion criteria were systemic pathology, previous hip surgery and body mass index > 32. In addition, potential sources of atypical motor behaviour in the lower

extremities had to be excluded; individuals were excluded when they affirmed an accident or surgery of either leg in the past year, recurrent knee or foot pain, concurrent low back and hip pain or use of medications which may affect the nervous system and reaction speed. Applicants with severe limitations were excluded by the requirements to be able to walk five steps up or down without holding a rail and to be able to lift the thigh to 90° hip flexion.

Suitable applicants were informed about the study (Appendices 8.5 and 8.6) and provided written consent (Appendix 8.7). For sample characterization and to separate subjects with symptoms that are atypical for hip joint pathology, subjects with pain were asked for their relevant medical history, they were examined physically (Appendix 8.8) and they completed the WOMAC (Bellamy et al 1988) and the Harris Hip Score(HHS) (Harris 1969) (Appendices 8.9 and 8.10).

8.2.2 Experiment and equipment

Participants performed six repetitions of step down from a step of 20 cm height onto a force plate (AMTI, Advanced Medical Technology, Watertown, USA) (Figure 8.2). Instructions were to stand parallel to the edge of the step and step down directly onto the force platform. Subjects were instructed to use the leg with the painful hip as leading limb. Controls were assigned to use the right or left leg for matching the subjects. Performance should be rapid and pain free.

Muscle activity of the superficial and deep gluteus medius and gluteus minimus was recorded by M-mode US (Xario XG, SSA-680A, Toshiba Pty Ltd, North Ryde, NSW, Australia) on the lateral hip of the leading limb (i.e. the painful hip). The US transducer (linear probe PLT-704SBT, 7.5 MHz, 3.8 cm footprint) was fixed on a custom-made triangular support that was strapped to the pelvis (Chapter Three, section 3.4.5.3.1). The recording location has been described by Ikezoe et al (2011), midway between the proximal end of the iliac crest and the greater trochanter. This simple, vertical recording location could be chosen because in Study Six no EMG electrodes had to be placed around the US transducer. A custom-programmed M-mode application was set with high contrasts, high frame-rate and highest sweep speed. The temporal resolution of the M-mode trace was 2.2 ms.

Ground reaction force of step-down was measured using a floor-inbuilt AMTI force plate (Advanced Mechanical Technology, Watertown, USA). Force data were sampled at 1000 Hz and collected together with kinematic data. Kinematic data were collected for monitoring task performance (VICON MX HD system, VICON Motion Systems, Oxford, UK, Nexus software, version1.6 with plug-in gait marker standard (http://wweb.uta.edu/faculty/ricard/Classes/KINE-5350/PIGManualver1.pdf). Kinematic data were sampled at 450 Hz.

US, force and kinematic data were synchronized by splitting the signal of a footswitch to go simultaneously into the ECG entrance of the US system to start the Vicon data collecting system. The ECG input of the Xario US system had been tested against the AUX input, which showed a less stable switch signal. The delay in the ECG channel, 2.2 ms, was minimal. The study protocol is documented in Appendix 8.11.

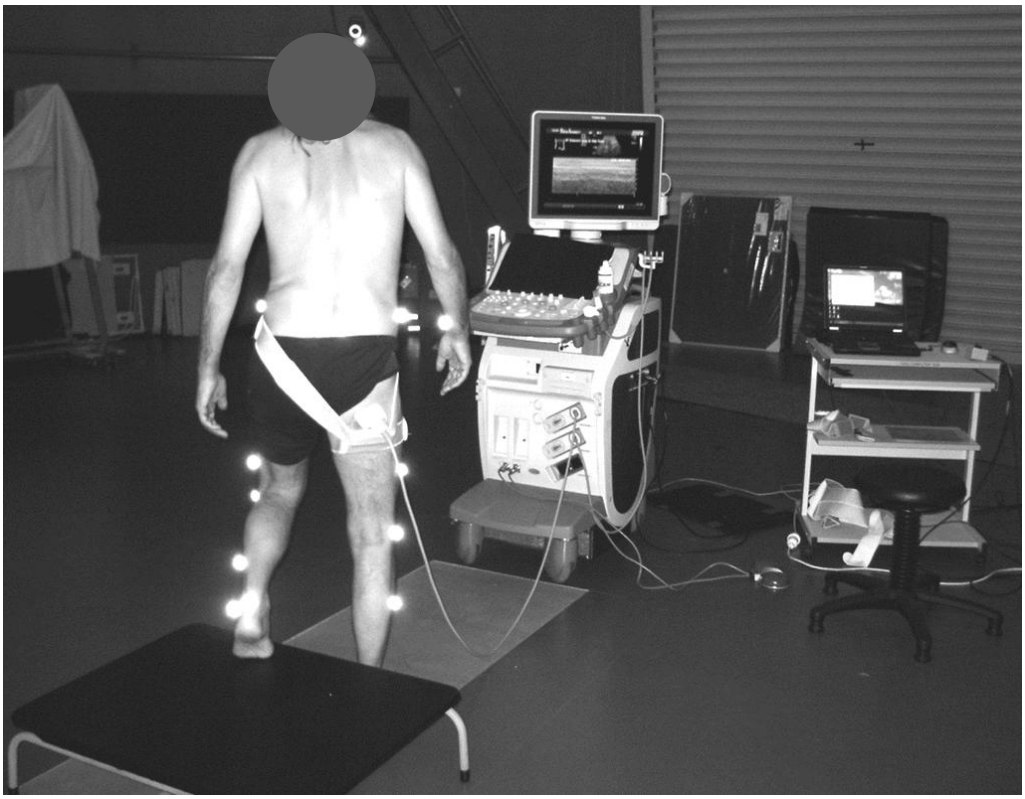


Figure 8.2. Subject performing a step down onto a floor-inbuilt force platform (US transducer here on the hip extensors for another investigation).

8.2.3 Data analysis

Onset of high-frequency M-mode patterns was determined for the superficial and deep gluteus medius and gluteus minimus using the computed detection method based on the Teager-Kaiser Energy Operator (TKEO) that has been described in Chapter Seven, sections 7.2 and 7.3. Briefly, the M-mode greyscale values of each pixel line in the depth of interest were extracted by a custom-written LabView program (LabVIEW V8.2.1, National Instruments, Texas) and transformed using the TKEO. The standard deviation (SD) of all TKEO lines in the depth of interest was used for recognition of high-frequency patterns. High-frequency patterns indicate rapid muscle tissue motion, and are influenced by fascicle diameter (Chapter Seven, section 7.4). Onset of high-frequency patterns was determined when the SD of the TKEO lines exceeded 800, 1200 or 2000 for 5 of 10 consecutive data points (discussion of the TKEO unit in Chapter Seven, section 7.2.2.4).

Onset of M-mode high-frequency patterns was related to ground reaction force. Computed detection with visual control was used to determine the instant of initial contact on the force platform when the signal exceeded the threshold value of 30 N. The instant of peak ground reaction force was determined by computer using the minimum function, and was controlled visually. As subjects with hip pain tend to perform lower limb tasks more slowly, the latencies between high-frequency pattern onset and peak ground reaction force were time-normalized to loading time. Loading time was calculated as the time span between initial contact and the instant of peak ground reaction force (Figure 8.4). The outcome measurement was the time-normalized latency between the onset of high-frequency M-mode patterns, SD800, SD1200 and SD2000, and the instant of peak ground reaction force for the three levels of depth of the hip abductor muscles, the superficial and deep gluteus medius and gluteus minimus.

Trials were included when step down was performed with the correct leg, loading time was between 50 and 250 ms, the synchronization signal was included on the M-mode trace and onset of M-mode high-frequency patterns could be detected.

8.2.4 Statistics

Data distribution was evaluated by visual inspection of histograms and boxplots and by the Shapiro-Wilk test for normal distribution. If the assumption of normality was met in both groups, parametric analysis was applied; otherwise non-parametric analysis

was used.
Loading times in each group were described by mean, standard deviation (SD) and by median. Group differences in loading times were examined by the Mann-Whitney test. In each group, the time-normalized latencies between onset of high-frequency patterns and peak ground reaction force were averaged per subject. Latencies were described per group by mean, SD and 95% confidence intervals, alternatively by median and interquartile range. Group differences in latencies were examined for each muscle level separately using the *t*-test or the Mann-Whitney test. Group differences were recalculated into ms based on the group-specific median of loading times.
Activity patterns of the hip abductors were visualized by the mean and confidence intervals of the time-normalized latencies separately for each muscle level. Significance was set to $\alpha = 0.05$. Results are reported as mean (SD) if not stated otherwise.

8.3 Results

More than 200 interested individuals were screened to recruit 35 subjects and 35 controls who were informed about the study and signed consent. As age is a major risk factor for degenerative joint disease (Suri et al 2012), controls were sampled to be age matched or older than the subjects. During recruitment, it became evident that the subjects with pain fell into two subgroups. One subgroup consisted of predominantly male participants with a medical diagnosis of hip OA who were approximately 60 years of age and older. The second subgroup consisted of predominantly female participants without OA diagnosis or clear signs of OA. The second subgroup was younger, in average approximately 50 years. Controls were recruited to match the subgroups to enable a subgroup comparison. Participants' and subgroups' characteristics are reported in Table 8.1.

In each group, 210 trials were recorded of which 203 pain trials (97%) and 197 control trials (94%) resulted in complete datasets for at least one M-mode variable. Variable detection rates were excellent (Table 8.2).

Table 8.1. Participants' and subgroups' characteristics; fewer participants than those included in the main groups were allocated to subgroups in order to keep subgroups homogenous.

	Age, years	BMI, kg/m²	females	males	Symptom duration, years	WOMAC	HHS	VAS bad hip pain day
Subjects with pain	53.8	26.4	20	15	6.9	65.5	72.9	5.6
OA subgroup	57.9	26.9	4	10	6.4	69.1	66.9	5.2
Young pain subgroup	46.2	25.8	9	0	6.3	39.4	81.8	4.8
Controls	56.6	25.1	22	13				
OA-control subgroup	60.9	26.3	4	10				
Young control subgroup	49.3	24.4	9	0				

Table 8.2. Variable detection rates

	Superficial gluteus medius			Deep gluteus medius			Gluteus minimus		
	SD800	SD1200	SD2000	SD800	SD1200	SD2000	SD800	SD1200	SD2000
Pain	99.5%	97.9%	97.0%	99.0%	100%	100%	75.8%	92.5%	100%
Control	99.5%	99.5%	99.5%	99.0%	99.5%	100%	86.9%	96.9%	100%

In both groups, loading times were distributed significantly skewed to the right. Subjects with hip pain had a loading time of 126 (31) ms, median was 120 ms. Control subjects had a loading time of 117 (24) ms, median was 113 ms. Loading time was significantly longer in subjects with pain, p 0.007.

For most variables, the distribution of latencies between high-frequency pattern onset and instant of peak force were close to a normal, with one exception. The Shapiro-Wilk test indicated a significant aberration from normal distribution for the SD800 variable measured on the deep gluteus medius in the control group, p 0.046.

Means and medians of the latencies indicate a general trend towards an earlier start of high-frequency pattern in subjects with pain (Table 8.3 and Figure 8.3). In gluteus minimus, high-frequency pattern were detected significantly earlier in subjects with hip pain, 103 ms using the SD1200 and 78 ms using the SD2000 variable (Table 8.4 and Figure 8.3). While in control subjects, onsets of high-frequency patterns in the deep gluteus medius and gluteus minimus were close, a significantly larger difference of gluteus medius and minimus onset was observed in subjects with pain, p 0.017 (Figure 8.3).

Table 8.3.Time-normalized and for each subject averaged latencies between onset of M-mode high-frequency pattern (variables SD800, SD1200 and SD2000) and instant of peak ground reaction force. + latency, M-mode onset was detected before peak ground reaction force. CI, 95% confidence interval, IQR, interquartile range. Parametric descriptors in non-normally distributed data in grey font.

Median (IQR), Mean ±CI	SD800		SD1200		SD2000	
	Pain	Control	Pain	Control	Pain	Control
Superficial gluteus medius	*130 (122*) 133 ±43	*91 (161)* 91 ±46	*76 (212)* 74 ±46	*19 (158)* 21 ±37	*5 (128)* 6 ±41	*-38 (171)* -36 ±33
Deep gluteus medius	*238 (179)* 223 ±44	*186 (119)* 195 ±34	*180 (180)* 167 ±40	*141 (151)* 134 ±35	*111 (190)* 84 ±40	*80 (171)* 63 ±39
Gluteus minimus	*308 (183)* 295 ±49	*233 (146)* 266 ±50	*262 (117)* 246 ±36	*170 (91)* 170 ±31	*145 (175)* 140 ±40	*84 (132)* 80 ±37

Table 8.4. Group differences of time-normalized and for each subject averaged latencies between M-mode high-frequency pattern onset (SD800, SD1200 and SD2000) and instant of peak ground reaction force; group difference recalculated into ms. *significant at 0.05.

Mean time-normalized difference ≙ ms; *p*	SD800	SD1200	SD2000
Superficial gluteus medius	42 ≙ 57 ms; 0.226	53 ≙ 65 ms; 0.070	42 ≙ 48 ms; 0.112
Deep gluteus medius (*non-parametric analysis in italics)*	*52 ≙ 75 ms; 0.067*[1]	33 ≙ 49 ms; 0.212	21 ≙ 30 ms; 0.438
Gluteus minimus	29 ≙ 53 ms; 0.388	76 ≙ 103 ms; 0.002*	60 ≙ 78 ms; 0.028*

[1] *Mann-Whitney test*.

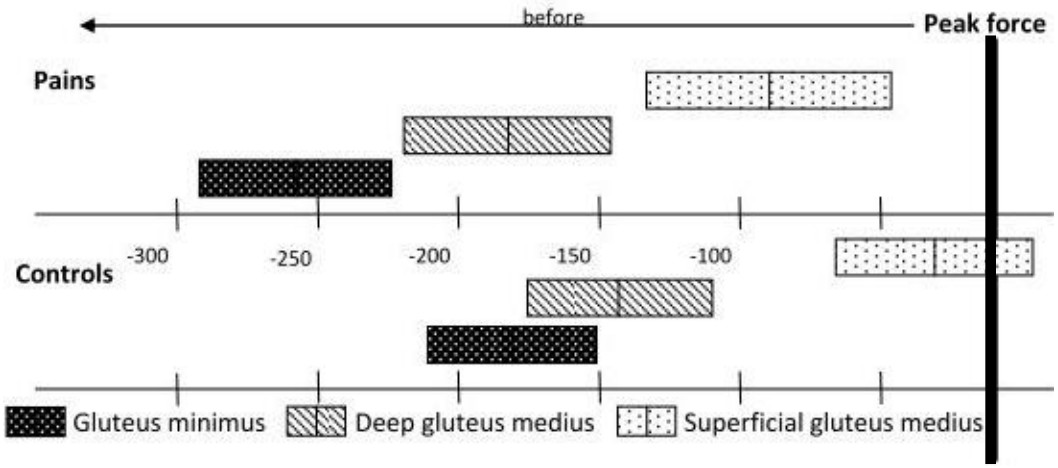


Figure 8.3. For subjects with hip pain, mean ± CI of onset of high-frequency M-mode pattern (SD1200) relative to peak ground reaction force (vertical, black line) indicate a trend towards earlier occurence of high-frequency patterns in all three abductor levels. In gluteus minimus, a significantly earlier occurrence of high-frequency pattern was detected. Controls (n=35), subjects with hip pain (n=35), time-scale refers to time-normalized data: percentage of loading time.

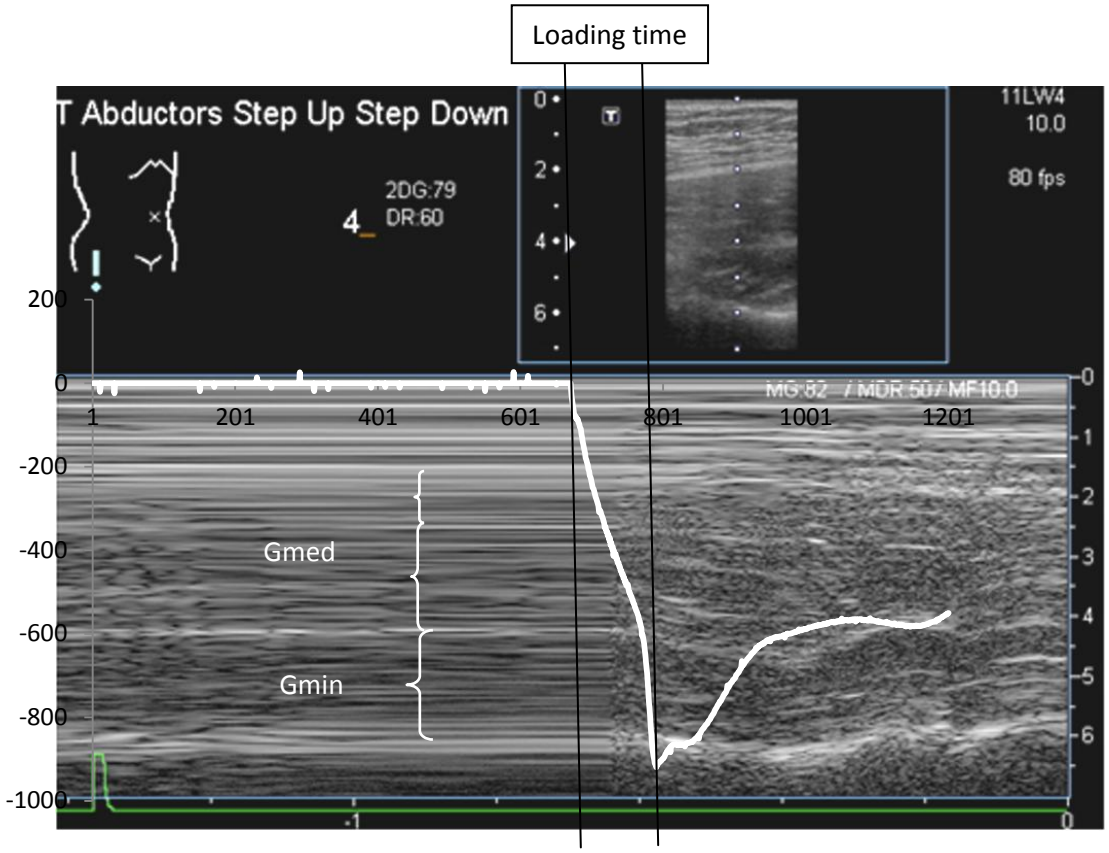


Figure 8.4. Example of a control trial, M-mode trace with superimposed signal of ground reaction force: high-frequency pattern start with the increase of ground reaction force. Loading time was defined as the interval between initial contact on the force plate and peak ground reaction force.

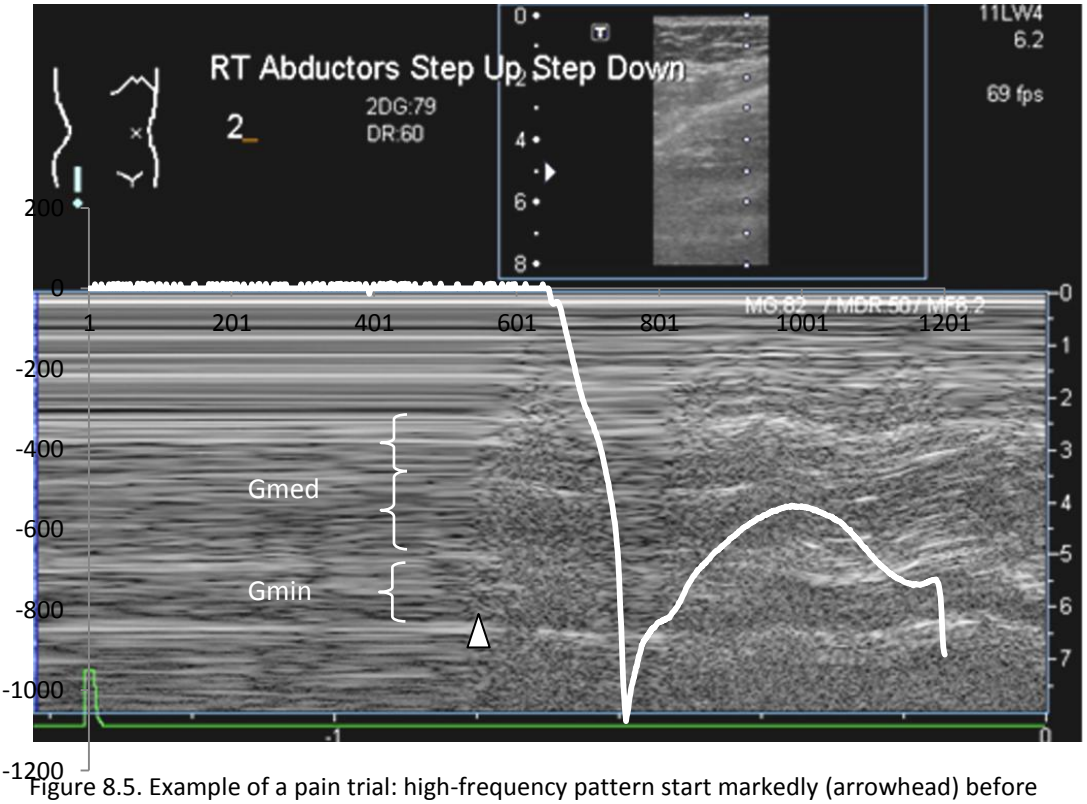


Figure 8.5. Example of a pain trial: high-frequency pattern start markedly (arrowhead) before initial contact on the force plate. Note also a lack of relaxation before onset.

No significant differences between the subgroups with pain were found. Clinically, both subgroups with pain demonstrated positive Inner Quadrant / FADDIR tests (Martin et al 2010, Martin and Sekiya 2008) in all / all but one subject. The Faber test (Martin and Sekiya 2008) and 'pain with active hip flexion' were positive in the majority of subjects in both subgroups. The older OA subgroup was more limited in range of motion than the younger subgroup. Both subgroups with pain demonstrated earlier onset of high-frequency M-mode patterns. The strongest difference of the onsets of deep gluteus medius and gluteus minimus motion was detectable in the younger subjects without OA. All subgroups demonstrated a trend towards earlier onset of high-frequency M-mode pattern with age (Figures 8.6 and 8.7).

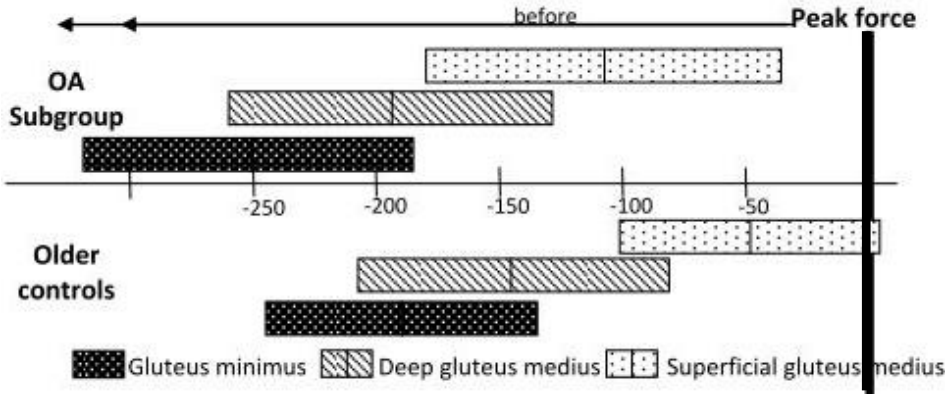


Figure 8.6. For subjects with hip osteoarthritis (n=14), mean ± CI of onset of high-frequency M-mode pattern (SD1200) relative to peak ground reaction force (vertical, black line) indicate a trend towards earlier occurence of high-frequency patterns in all three abductor levels. Note earlier onset of high-frequency pattern in the older control subjects (n=14) compared to the younger groups in Figures 8.3 and 8.7. Time-scale refers to time-normalized data: percentage of loading time.

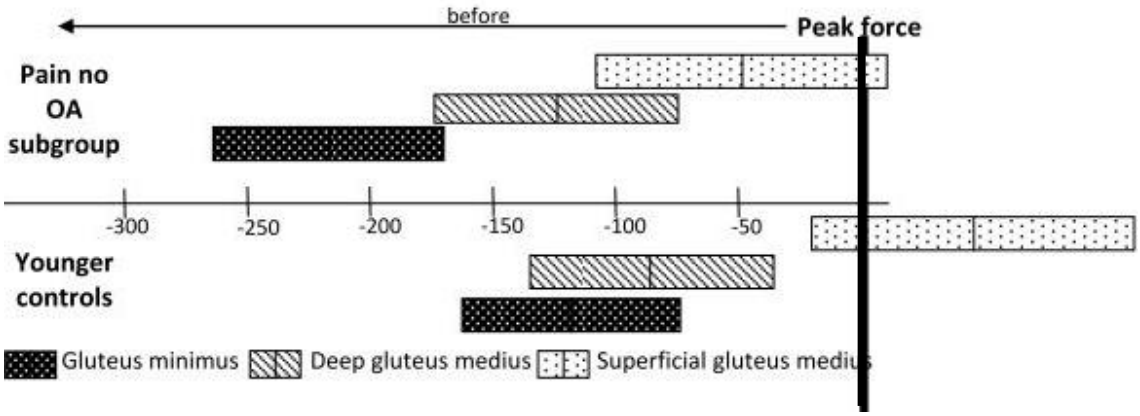


Figure 8.7. For female subjects with pain without OA signs (n=9), mean ± CI of onset of high-frequency M-mode pattern (SD1200) relative to peak ground reaction force (vertical, black line) indicate a trend towards earlier occurence of high-frequency patterns in all three abductor levels, significant in gluteus minimus. Time-scale: percentage of loading time.

8.4 Discussion

Based on M-mode US imaging, Study Six examined the motor activity of the two main hip abductor muscles during step down. M-mode high-frequency patterns demonstrated significant differences for gluteus minimus in subjects with hip pain and controls. The onset of high-frequency greyscale patterns relative to peak ground reaction force was significantly earlier in the gluteus minimus in subjects with non-traumatic, anterior hip pain. Earlier gluteus minimus motion was particularly pronounced in the younger subgroup of females with hip pain but without signs of OA. None of the clinical tests or assessments was explanatory for the larger difference between deep gluteus medius and minimus muscle motion in the younger, 'only pain' subgroup. The older subgroup with a diagnosis of hip OA stated more pain on VAS, had a lower Harris Hip Score (= more physical restrictions) and a higher WOMAC (= more subjective limitations), but did not report a longer symptom duration than the younger pain subgroup.

The latencies between high-frequency patterns and peak ground reaction force were highly variable, higher than in the isometric abduction task in Study Five (Appendix 7.1). Variation was similar between groups. In spite of the instruction to perform the step down rapidly, the study results document significant group differences in task performance. A longer loading time indicates a slower step down in the pain group. Also, differences in the shape of the force signal reflect group differences in task performance (Figures 8.4 and 8.5). High variation of motor performance in a step down task has been reported in the literature (Cavazzuti et al 2010) and may be interpreted as an indication of the high adaptability of the motor system to varying conditions, such as pain, in order to perform elementary tasks.

8.4.1 What is measured by M-mode high-frequency patterns?

High-frequency M-mode patterns measure a combination of two determinants of greyscale frequency, the speed of muscle motion and fascicle diameter. Muscle motion is a dynamic parameter and fascicle diameter a structural, static parameter. Muscle motion and fascicle diameter are both related to the use of a muscle. With less muscle activity, fascicle diameter decreases, the muscle shows atrophy.

Gluteus minimus is by its function and composition at a higher risk for muscle atrophy. Lieber (2010, p. 185 and p. 189) stated that antigravity muscles that cross only one joint and that are mainly composed of slow twitch fibres atrophy to a greater extent by decreased use than others. In the course of hip joint degeneration, atrophy of fast twitch fibres has also been reported. A large muscle biopsy study (Sĭrca and Susec-Michieli 1980) on the gluteus maximus, gluteus medius and tensor fasciae latae muscles of subjects with hip OA and controls demonstrated that selective atrophy of fast twitch fibres was significantly higher in the OA group. Both groups exhibited diminishing muscle fibre diameter of all fibres and a decrease of the relative number of fast twitch fibres with age (Sĭrca and Susec-Michieli 1980). Fibre atrophy may explain the general trends towards earlier high-frequency pattern detection with age and with pathology in the study results. Apart from these general trends, the study results indicate a difference in the temporal relationship of the onsets of high-frequency patterns in the deep gluteus medius and minimus muscles that points to a specific effect of hip pain on gluteus minimus.

The methodology used in Study Six does not allow for distinguishing the dynamic and the static components of M-mode high-frequency measurements. The proposed interpretation of the study result is therefore: M-mode high-frequency patterns indicate an altered use of gluteus minimus in subjects with hip pain.

Gluteus minimus motion differed between groups in the phase directly before initial contact, a phase when the limb has left its support base on the step and was moved through the air while preparing for the impact of imminent loading. This preparation phase differed between groups. Control subjects started activity shortly before the impact, just before the activity was needed. It appears that subjects with pain had either problems to relax and then restart activity, or that they applied a preventive strategy of early activity. A need for caution was also expressed in the longer loading time of subjects with pain.

8.4.2 Limitations

The effects of age and gender cannot be distinguished in this comparison due to gender differences between subgroups. Most females who applied and would have

suited the OA subgroup could not be included because of concurrent lower limb or low back pain. Further studies require a more homogenous sample which requires a narrower definition of diagnosis and age of inclusion.

The application of M-mode high-frequency patterns in the methodology used in this study did not allow for the differentiation of dynamic and static components. However, it needs to be known whether motion onset would be significantly different between groups when the measurement is adjusted for fascicle size. Differentiation was not feasible because the same threshold values were used for all three muscle levels and all individuals. Theoretically, a separation of the dynamic and the static component in M-mode high-frequency measurements should be possible by a normalization procedure, comparable to the MVIC normalization. It is necessary to determine an individual, muscle-specific threshold to adjust for structural differences.

The mixture of dynamic and static components in US measurements is typical for US imaging. Displacement, the feature measured by US imaging, is determined by the energy input relative to mechanical constraints, which are predominantly static parameters.

A further exploration and quantification of the mechanical constraints is necessary to understand the interplay of muscle activation and connective tissue properties and to isolate dynamic components. Study Six presents a starting point with the demonstration that M-mode delineates clinically relevant differences in deep muscle activity.

8.4.3 Implications

Study Six indicates functional differences between the gluteus medius and minimus muscles, and also between the superficial and the deep gluteus medius muscle levels. To the author's knowledge, Study Six is the first study that indicates a primary influence of hip pathology on gluteus minimus; and it implies the need for specific exercise to target gluteus minimus. Kumagai et al (1997) compared the response of gluteus medius and minimus to different exercises. This Japanese MRI study reported particularly high metabolic activity in gluteus minimus following isometric hip abduction in 20° of hip abduction and following one-legged stance. Most literature on hip

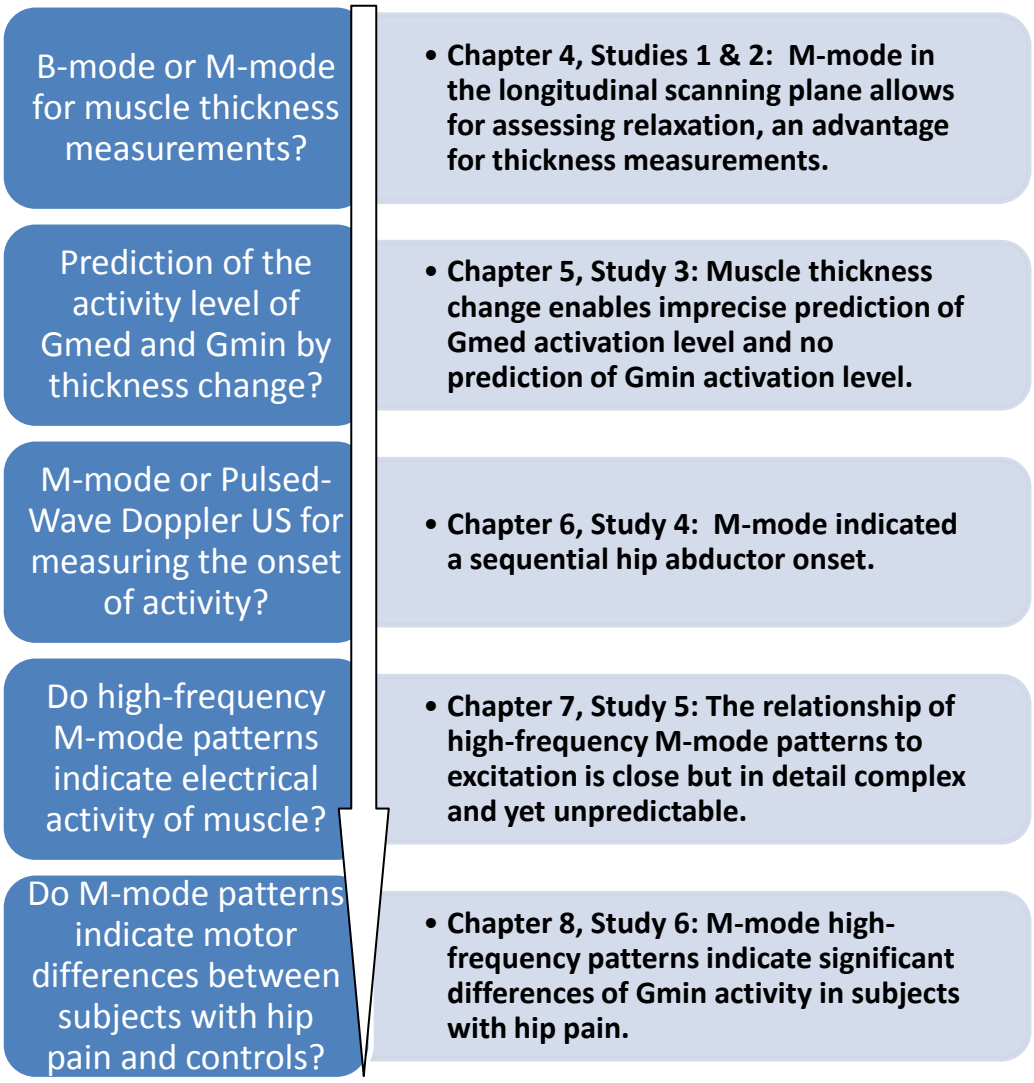
abductor exercises is limited to the superficially accessible muscles gluteus medius, tensor fascia latae and gluteus maximus (Selkowitz et al 2013, Willcox and Burden 2013, Cambridge et al 2012). Further research is required to identify exercises which target gluteus minimus.

Study Six indicates the clinical relevance of M-mode high-frequency pattern measurements. One option to detect and compute greyscale patterns has been suggested. More research is needed to reappraise and scrutinize the proposed approach. This final study of the thesis provided a starting point.

M-mode US imaging enables an insight into deep muscle activity and provides the means to differentiate synergistic muscle activity by its mechanical aspect. The appraisal of the effects of such differentiated therapy on hip pain and on the development of hip joint degeneration may lead to innovative physiotherapeutic treatment.

8.4.4 Acknowledgements

I acknowledge the advice and support by Toshiba Australia, Medical Division and the expert help by Curtise Ng from the Department of Medical Imaging in DICOM image transfer. I want to thank Rhonda Jackson from Curtin FM for the intense support of subject recruitment, and I thank the subjects for making this study possible.


B-mode or M-mode for muscle thickness measurements?
• Chapter 4, Studies 1 & 2: M-mode in the longitudinal scanning plane allows for assessing relaxation, an advantage for thickness measurements.
Prediction of the activity level of Gmed and Gmin by thickness change?
• Chapter 5, Study 3: Muscle thickness change enables imprecise prediction of Gmed activation level and no prediction of Gmin activation level.
M-mode or Pulsed-Wave Doppler US for measuring the onset of activity?
• Chapter 6, Study 4: M-mode indicated a sequential hip abductor onset.
Do high-frequency M-mode patterns indicate electrical activity of muscle?
• Chapter 7, Study 5: The relationship of high-frequency M-mode patterns to excitation is close but in detail complex and yet unpredictable.
Do M-mode patterns indicate motor differences between subjects with hip pain and controls?
• Chapter 8, Study 6: M-mode high-frequency patterns indicate significant differences of Gmin activity in subjects with hip pain.

9 Chapter Nine: Discussion and Conclusions: M-Mode US Demonstrates Altered Hip Abductor Activity in Individuals with Hip Pain

Overview of Chapter Nine

- Section 9.1 gives a short introduction.
- Section 9.2 reviews the findings of the presented research process.
- Section 9.3.discusses the findings and their limitations in the context of current literature.
- Section 9.4 discusses the clinical implications of the research findings and proposes directions for further research.
- Section 9.5 states the conclusions.

9.1 Introduction

Non-traumatic, anterior hip pain in middle or older age is an indicator of an increased risk to develop hip joint degeneration and is a health problem with growing relevance in ageing societies. Physiotherapists are challenged to explore conservative treatment options, develop targeted interventions and provide evidence for their efficacy. An important obstacle for answering the challenge is the very limited insight into in-vivo activity patterns of deep hip muscles and their alterations with pain. Current methods to assess selective muscle activity are either limited to superficially accessible muscles or invasive and clinically not applicable methods. Driven by the clinical need to explore the effects of hip pain on the in-vivo coordination of superficial and deep hip muscles, gluteus medius and minimus were chosen as clinically relevant muscle models for developing US measurement methodology. The series of studies presented in this thesis started with the application of the broadly applied US measurement of muscle thickness and continued to the proposition of new measurement variables of muscle motion and their application in a clinical study. This final chapter summarizes the research findings and discusses their value in the context of current literature. Considering the research limitations, implications and recommendations for future research are stated before the final conclusions are drawn.

9.2 Summary of the research findings

This research identified M-mode ultrasound (US) to be of specific interest for the evaluation of muscle activity. M-mode enables two major measurements of activating muscle, measurements of thickness change and of muscle motion. This combination makes M-mode US specifically attractive for the assessment of muscle activity. In comparison to B-mode US, the advantage of M-mode is the much higher temporal resolution and the ability to observe the process of activation in an image, a trace of motion over time. Alternative modes for motion measurements are Doppler modes. Doppler US modes do not enable a comparably good anatomical mapping of motion information. Doppler did not enable a comparison between different muscle levels. The major disadvantage of Doppler based information is the dependency on the

angle of motion relative to the US transducer. Different components of muscle motion are biased depending on the position of the muscle relative to the transducer.

The application of M-mode US to assess the activity of the gluteus medius and minimus muscles provided several, new findings. Study Three demonstrated a poor applicability of *thickness measurements* for determining the level of gluteus medius and minimus activation. The level of gluteus medius activity could be roughly estimated by the increase of muscle thickness; however, surface EMG enables more precise estimates also non-invasively, and is therefore of higher clinical relevance. The level of gluteus minimus activity could not be estimated from thickness change, as the thickness of gluteus minimus did not change with activity. This finding was interpreted with regard to the change of architectural parameters during activation, an approach underrepresented in current literature.

M-mode *motion measurements* provided an insight into the sequence of activity of the hip abductors. The assessment of the onsets of gluteus medius and minimus motion in Study Four demonstrated an earlier onset of gluteus minimus motion by 36 ms. Study Five confirmed the earlier onset by 34 ms earlier gluteus minimus excitation. Earlier gluteus minimus activation supports the understanding that gluteus minimus has a more stabilizing role in hip joint control than gluteus medius. Sequential gluteus medius motion onsets indicated a functional division of gluteus medius into a superficial and a deep muscle level.

Study Four emphasized the *technical demands* of measurements of muscle motion. Motion measurements require the identification and control of accessory sources of motion, optimal data synchronization and high-resolution data storage. Motion that is not related to the activation process, such as transducer motion or undesired motion components in the performance of the task need to be controlled and minimized.

A complication of interpreting US measured muscle motion is that the source of the moving energy may be unclear. Activity-related motion does not necessarily indicate the concurrent, local electrical excitation of muscle. In Study Five, it was proposed to overcome this limitation by the introduction of *M-mode high-frequency greyscale patterns*, which occur with rapid muscle motion. High-frequency patterns can be detected visually and also by computer. The comparison of fine-wire EMG and M-

mode US data indicated a close relationship and high prediction of electrical muscle excitation by M-mode high-frequency variables. However, Study Five indicated also some yet unexplainable exceptions. In particular, the finding of high-frequency pattern onsets preceding gluteus minimus excitation was unexpected. According to the literature, the onset of motion should follow excitation in the first activating muscle. It can only be speculated that motion was transmitted from adjacent motor units which were not included in the fine-wire sample or from an adjacent muscle part that activated earlier out of the imaging plane. Further research on the translation of excitation into muscle contraction, on muscle-internal motion transmission and on the functional relevance of muscle motion in the physiology of muscle activation is necessary to explain the observed phenomena.

The results of Study Five suggested differences in the onset of M-mode high-frequency patterns between subjects with and without hip pain. This preliminary observation was confirmed in Study Six in a case –control design with 35 subjects with hip pain and 35 controls. The use of US measurements allowed for a sample size that is not achievable in fine-wire EMG studies. Study Six demonstrated a significant effect of hip pain on gluteus minimus activity but much less on gluteus medius.

Three different scanning sites have been used in the course of this research (Figure 5.1. p. 100, section 8.2.2 p. 182). The anterior oblique site (Studies One – Four) has been chosen for comparative measurements by surface EMG. The scanning location at Hochstetter's triangle (Studies Four and Five) allowed for safe comparative measurements by fine-wire EMG. The sagittal scanning site used in Study Six has been recommended in the literature (Ikezoe et al 2011) and is simple to reproduce. No scanning site provided the advantage of visualizing bony landmarks that would allow for visual control of the precise repositioning of the transducer from the image itself.

Table 9.1. Studies in this research

Study One	Chapter Four	Thickness measurements of gluteus medius and minimus: B-mode or M-mode ultrasound?	Reliability was high in both modes. M-mode enables identification of muscle relaxation, also in deep muscles.
Study Two	Chapter Four	Thickness measurements of hip abductor activity by M-mode ultrasound imaging: longitudinal or transverse scanning plane?	Thickness measurements on the longitudinally scanned hip abductors are preferable due to less transducer motion by muscle bulging.
Study Three	Chapter Five	Does ultrasound measured change of muscle thickness indicate the level of individual hip abductor activity?	Muscle thickness change enables imprecise prediction of gluteus medius and no prediction of gluteus minimus activation level.
Study Four	Chapter Six	Onset of gluteus medius activation measured by electromyography and ultrasound: M-mode or Pulsed Wave Doppler?	M-mode US indicated a sequential onset of activity-related motion in gluteus minimus and gluteus medius. The relationship between the onset of excitation and of motion is variable.
Study Five	Chapter Seven	M-mode measurements of the onset of deep muscle activity: are high-frequency patterns associated to muscle excitation?	M-mode high-frequency pattern are more closely related to muscle excitation than simple motion onsets.
Study Six	Chapter Eight	Do M-mode high-frequency greyscale patterns indicate altered hip abductor activity in subjects with hip pain?	Onset of M-mode high-frequency pattern in gluteus minimus is significantly earlier in subjects with hip pain.

9.3 The research findings in the context of current literature on ultrasound imaging of muscle activity

9.3.1 M-mode ultrasound

In this research, M-mode US was identified as specifically suited for the assessment of muscle activity. Current literature on US assessed muscle activity is dominated by B-mode US applications. B-mode is the most commonly used US mode which provides information that is closely related to sectional anatomy. B-mode US appears to be easy to understand as muscles can be observed in real-time 'at work'. The unexpected complexity of the interpretation of B-mode US recorded trunk and pelvic floor muscle activity has been described comprehensively by Whittaker (2007). A

minority of researchers applied M-mode US to measure abdominal muscle thickness change (Mannion et al 2008c, John and Beith 2007, McMeeken et al 2004) or onset of muscle motion (Unsgaard-Tondel et al 2012, Vasseljen et al 2012, Westad et al 2010, Vasseljen et al 2009, Vasseljen et al 2006). All cited studies were performed on the abdominal muscles except the study from 2006 in which the multifidus muscle was scanned. Also Mannion et al (2012) used M-mode for abdominal thickness and motion measurements, but determined motion onset by Colour Doppler information superimposed on the M-mode trace. The application of M-mode US in this research needed to be explorative because of the sparse literature.

The limitations of this exploration of M-mode US for rehabilitation are multiple. The investigation of artefacts was limited and needs extension. The comparison of EMG and M-mode measurements was restricted to activity onsets. Broadening the investigation to the duration and endpoint of activity would enhance the understanding of the relationship of electrical and mechanical aspects of activity reflected by the two modalities. The explorative approach including a range of force levels increased variation and weakened the strength of the study results. The statistical results would have been more definite if the studies had been performed in rapid and strong activity, comparable to the typical armlift tasks in motor control studies.

The rapid armlift task in EMG studies is a provocation that reveals differences in motor control that are less clear in a more 'natural' task performance (Hodges and Richardson 1999). In this research, it has been assumed that the 'falling' body weight during step down in Study Six constitutes a rapid provocation task comparable to the armlift task. In contrast to the armlift task, in which the speed of task performance was equal between groups or more rapid in the subjects with pain (Hodges and Richardson, 1999), Study Six demonstrated a slightly but significantly slower speed of step down in the group with hip pain. It may be noted that in most studies using the armlift task, group differences in task performance have not been evaluated.

In Studies One to Five of the presented research, clinical interests for the applicability of US measurements under less restricted conditions led to a study design including different force levels and a relatively self-selected speed of task performance. The resulting experience pointed to the difficulty to interpret slow muscle motion and led to the development of the high-frequency pattern variables.

The application of US modes other than B-mode requires work and time to understand the information provided by the mode and the sources of misinterpretation. To establish the use of other US modes, potential users need to be convinced that the information provided in this innovative application is clinically important. This research accomplished an investigation of M-mode applications beyond existing reports by providing initial evidence for a clinical benefit of using M-mode for measurement of muscle activity.

9.3.2 Muscle thickness measurements by ultrasound imaging

Study Three identified strong limitations in the application of thickness measurements on the hip abductors. While thickness measurements are the most often reported US measurements in rehabilitation research, the clinical value of thickness measurements of muscle activity remains controversial. Preferential regression models for predicting the level of muscle activity differed (McMeeken et al 2004, Hodges and Moseley 2003). Prediction of the force level by thickness change was not feasible in some muscles (Brown and McGill 2010, Delaney et al 2010, John and Beith 2007, Narici et al 1996). Some authors could demonstrate altered muscle activity in subjects with low back pain by US thickness measurements (Wallwork et al 2009, Kiesel et al 2007c, Ferreira et al 2004). Others failed to demonstrate group differences by muscle thickness for low back pain, neck pain or pelvic girdle pain (Peolsson et al 2012, Stuge et al 2012, Gubler et al 2010). With regard to therapeutic interventions, US thickness measurements indicated an immediate effect of spinal or cervical mobilization (Jesus-Moraleida et al 2011, Konitzer et al 2011, Koppenhaver et al 2011, Puentedura et al 2011, Kiesel et al 2007a) but failed to detect effects of prescribed exercises (Vasseljen and Fladmark 2010, Jansen et al 2009) or to document changes in motor activity with relevance to the clinical outcome (Mannion et al 2012, Unsgaard-Tondel et al 2012).

An important limitation of thickness measurements by US is the measurement variation which decreases precision and responsiveness to change (Teyhen et al 2011, Koppenhaver et al 2009b). High variation was a limitation of Studies One to Three on abductor thickness change in this thesis. The variation of US thickness measurements has multiple sources, some of which are related to technical aspects, some to the handling of the measurement process and some to the influence of yet insufficiently

determined confounders. All these sources of variation reflect a potential for improvement.

Accurate measurements require not just a basic image but the optimal achievable US image. A consensus on transducer handling and training for setting the US system is required to improve a consistent acquisition of high quality images. Quality standards for measurement acquisition need consensus and dissemination (Ishida and Watanabe 2012, Whittaker et al 2010, Whittaker et al 2009). Education and training, together with the adaption of the 'hardware' for the specific requirements of imaging muscle activity, are the most important aspects to improve the technique of measurements by US. For example, the extension of the footprint, beam steering in B- and M-mode, an indication of transducer tilt and options for transducer fixation need consideration.

The interpretation of thickness change of a muscle requires knowledge of the muscle-specific relationship of thickness change and increase of the activity level under isometric conditions. However, under dynamic conditions thickness change is an indicator of muscle length, not force level (Shi et al 2009, Xie et al 2009, Herbert and Gandevia 1995). Any change in muscle length is a confounder in the estimation of the activity level from muscle thickness change. Differences in muscle thickness in different postures or exercises may reflect a combination of activity and muscle length changes and are therefore difficult to interpret (Brown and McGill 2008).

As discussed in Chapter Five, the arrangement and dynamic behaviour of muscle fascicles likely constitute important determinants of muscle thickness change during activity (Rudroff et al 2008, Narici et al 1996). US imaging reflects muscle activity as the effect of electrical excitation on the mechanical constitution of the muscle within environmental constraints. This statement may be illustrated by the comparison to a motor. Electrical excitation is the ignition, muscle metabolism the fuel. The architectural organisation and connective tissue structures are the mechanics that translate the power of the motor into a directed force. One needs to understand the regular mechanics to be able to detect aberrations in work.

9.3.3 Muscle motion measurements by ultrasound

This research focussed on M-mode motion measurements, as thickness measurements were not suited to answer the main research question for differences in motor patterns with hip pain. The observation and measurement of the onsets of hip abductor activity-related motion supported the division of the gluteus medius muscle into different parts (Gottschalk et al 1989) and added the functional distinction of two levels of depth.

The literature on motion measurements of muscle activity by M-mode is sparse (Westad et al 2010, Vasseljen et al 2009, Vasseljen et al 2006). Motion onset has been measured also using Tissue Doppler Imaging (TDI). These studies are included into the discussion because they extend the scope of the achievements of onset measurements (Mannion et al 2012, Gubler et al 2010, Mannion et al 2008a, Pulkovski et al 2008). In a case-control study including 48 subjects with chronic low back pain, Tissue Doppler US measured onset of abdominal muscle motion indicated anticipative activity in both groups and a significantly earlier onset of right abdominal activity in the pain group (Gubler et al 2010). The study result challenged evidence for a delayed onset of transversus abdominis activity with chronic low back pain from smaller fine-wire EMG studies (Hodges and Richardson 1998, Hodges and Richardson 1996). An M-mode US study on 109 volunteers with chronic low back pain detected no change in the onset of abdominal muscle activity after an eight-week core stability training focussing on isolated deep trunk and pelvic floor muscle activity (Vasseljen et al 2012). The study result questioned the finding of persisting improved feedforward activation of transversus abdominis following a four-week training of isolated activation, which has been reported from a fine-wire EMG study on nine subjects with chronic low back pain (Tsao and Hodges 2008). Study Six of this thesis demonstrated an early onset of gluteus minimus motion in subjects with hip pain instead of the expected delay.

Current knowledge of the relationship between excitation and muscle motion is still too limited to interpret the inconsistencies of US and EMG studies. The high variability in the relationship between EMG and US onsets is due to yet undetermined influences, and questions conclusions of excitation onset from motion onset. To the author's knowledge, the validation studies on which the above stated clinical studies are based, were performed on asymptomatic subjects (Vasseljen et al 2009, Mannion et al 2008a, Vasseljen et al 2006). The interpretation of US measurements on symptomatic

subjects relies on the assumption that method correlation would be consistent despite pathology. The results of Study Five challenge this assumption. A different relationship between excitation and muscle motion is reasonable because studies indicated that altered muscle activity induces changes of the structural properties of muscle (Barber et al 2011, Gao et al 2009, Falla and Farina 2005, Lieber et al 2004).

The main issue of motion measurements by US is the interpretation. Existing studies demonstrated that US detected muscle motion does not need to be associated with local electrical excitation (Westad et al 2010, Vasseljen et al 2009, Mannion et al 2008a, Vasseljen et al 2006). Muscle motion preceding excitation has been explained by transmission from an adjacent muscle (Vasseljen et al 2009, Mannion et al 2008a, Vasseljen et al 2006). The results of Study Five indicated limitations of this explanation. Only a minor part of early motion onsets was explainable by motion in the neighboured muscle. Other sources of motion were not investigated, in particular those out of the limited dimensions of the US scanning plane and the small sample of fine-wire electrodes. Theoretically, 3D ultrasound would be suited to visualize a thicker 'slice' of tissue and to broaden the field of view to adjacent sources of motion. Practically and at the current stage of technology, 3D ultrasound provides no alternative of interest for investigations of the timing of muscle activity. The 3D volumes are constructed from multiple 2D sections. Due to the high amount of processing work, the frame rate and hence the temporal resolution is low (Hoskins 2010, pp. 171-173).

An intriguing option to explain muscle motion without electrical excitation was raised by Allison (2012) who suggested that US detected onsets of muscle motion may arise from relaxation following low level, sustained activity. This suggestion implies that low level activity has not been identified. M-mode US enables a clear, visual distinction of low level activity and relaxation. A misinterpretation of muscle motion at the end of activity can be avoided and is not the source of unexplained muscle motion in Study Five. Further research is required to develop the understanding of activity-related muscle motion.

Chapter Seven reported a further step of this research, an innovative type of motion variable that can be detected by computer. The close relationship of M-mode high-frequency patterns to electrical excitation and the clinical relevance of the new motion variables were demonstrated in Studies Five and Six. A limitation of high-

frequency pattern measurements is that with the methodology used, the dynamic aspect (speed of muscle motion) could not be distinguished from the static aspect (fascicle diameter). This limitation may be overcome by use of a normalization procedure comparable to the test of maximal voluntary isometric contraction (MVIC). The presented research provided a step in the development of US measurements of muscle motion and their interpretation. Further research needs to follow.

9.3.4 Altered motor activity in non-traumatic, anterior hip pain measured by M-mode ultrasound

The results of Study Six demonstrated the largest effect of hip pain in gluteus minimus. Computer evaluation and visual data inspection indicated an earlier onset of abductor muscle motion relative to the impact of step down. Least change of activity was found in the deep gluteus medius, which is the thickest muscle bulk of the hip abductors. It appears that the timing of deep gluteus medius activity needs to be preserved to be able to step down. Altered timing of activity, but not significant, was also detected in the superficial gluteus medius muscle.

The finding of early gluteus minimus activity in subjects with pain was unexpected. Motor control theory suggests a delayed, but not an early activation onset in pain affected, deep muscles (Falla et al 2004a, Cowan et al 1999, Hodges and Richardson 1998). However, current literature is controversial about normal versus pain affected muscle behaviour. The concept of spinal stabilization by bilateral, symmetrical, directionally independent, feedforward activity of transversus abdominis has been questioned by a study using the Doppler based Strain-Rate US (Westad et al 2010) and by a fine-wire EMG study (Allison and Morris 2008). The concept of patella stabilization by an early onset of vastus medialis obliquus activity (Cowan et al 1999) has been questioned by a surface EMG study (Cavazzuti et al 2010). A large study by Tissue Doppler US demonstrated earlier motion onset of transversus abdominis in subjects with chronic low back pain, instead of the expected delay (Gubler et al 2010). The clinical relevance of a anticipative onset of transversus abdominis activity has been contested in two large US studies by Tissue Doppler US and M-mode that did not find a significant influence of the timing of abdominal muscle activity on any clinical measure of low back pain (Mannion et al 2012, Vasseljen et al 2012). Differences in the speed of

task performance, the small sample sizes of fine-wire studies or even an influence by the fine-wires themselves may play a role in inconsistent study results.

A recently published theory on motor adoptions to pain negates a simple, predictable alteration of motor patterns (Hodges and Tucker 2011). Instead, Hodges argues that "activity is redistributed within and between muscles rather than stereotypical inhibition or excitation of muscles" (Hodges 2011, p. 220). A central element of the theory is the individual modification of motor patterns with the aim to protect tissues from pain, with short-term benefits on the expense of potential, negative, long-term consequences, e.g. increased local joint loading and restrictions in range of movement and variability of motor behaviour (Hodges 2011, Hodges and Tucker 2011). Individually different effects of pain on motor patterns emphasize the need for a clinically applicable assessment of motor patterns in order to address the individually most affected muscles in physiotherapy.

9.4 Implications for future research

9.4.1 M-mode ultrasound and the mechanics of muscle activity

The main limitation of the results of Study Six is the missing differentiation between the dynamic and static determinants of M-mode high-frequency patterns. The next step should be a normalization procedure to account for individual and muscle specific differences in fascicle diameter in order to isolate the dynamic component in high-frequency pattern measurements.

From a methodological perspective, the development of US measurements of activity related motion needs an enhanced understanding of the mechanics of muscle tissue. Beyond the influence of muscle architecture, mechanical properties, in particular the elasticity of muscle tissue and mechanisms of motion transmission, need exploration.

Studies Five and Six suggest that the physiological process of excitation-contraction coupling is not the only source of local motion in activating muscle groups. Motion as well as force is an expression of energy which is transmitted within muscle. The role of the connective tissue in force transmission within and between muscles has

been increasingly acknowledged in the literature (Gillies and Lieber 2011, Maas and Sandercock 2010, Purslow 2010, Haraldsson et al 2008, Huijing 2007, Yucesoy and Huijing 2007, Hatze 2002, Sheard and Duxson 2002, Huijing 1999). It has been shown that extracellular matrix components are mechanosensitive. Mechanical stimuli influence cell metabolism, chemical cell interaction, growth and mechanical tissue properties (Chiquet et al 2009, Janmey et al 2009, Lele and Kumar 2007, Grounds et al 2005). Muscle is a sensor of the function imposed on it and it adapts immediately to changed demands (Lieber 2010).

US imaging reveals mechanical aspects of muscle activity. Currently, mechanical activity is difficult to interpret because the concept of muscle activity is defined by the electrical state of activation. Further research into the translation of excitation into mechanical response may broaden the current understanding of muscle activity.

9.4.2 Clinical utility of M-mode ultrasound application

M-mode traces contain readily accessible information on muscle relaxation and activity-related muscle motion which can be observed without complicated computed pattern detection. Muscle relaxation and disturbances of relaxation can be easily recognized (Figures 4.4 and 4.5). Different qualities of motion can be distinguished by observation (Figure 7.1). Sequential motion onsets and differences in the intensity of motion with depth are indicated clearly (Figures 7.14 and 7.15).

Before these observations can be interpreted regarding muscle dysfunction, normal patterns of activity-related motion and their normal variation needs to be known. Descriptive studies which provide reference of the range of normal activity patterns are needed. The high variation found in Study Six indicates that several repetitions of a task need to be observed before conclusions on disturbed gluteus minimus function can be drawn. However, in single cases, altered patterns of activity may be unambiguous and impressive, e.g. avoidance to use gluteus medius in Figure 7.16.

9.4.3 Therapy of altered motor activity in non-traumatic, anterior hip pain

Study Six assessed hip abductor activity in an activity of daily living, a step down and found significant alterations in the use of gluteus minimus. The treatment of

gluteus minimus by exercise and the assessment of the influence of the intervention on the patterns of activity and pain would be a logical progression of the research. Which exercise is suitable to address gluteus minimus? M-mode may be used to evaluate typical hip abductor exercises and identify exercises which have a particular effect on gluteus minimus activity patterns. The vision behind these practical suggestions is an US guided re-training of individual, pain-related motor dysfunction.

According to Cram and Kasman (1998), approximately 90% of the human skeletal muscles are not assessable by surface EMG. Most research on muscle activity relates to the 10% of muscles that are accessible. The assessment, the understanding and the therapy of deep muscle activity bears a great potential for improved and more specific physiotherapy.

9.5 Final conclusions

M-mode US enables thickness and motion measurements of muscle activity. The measurement of the change of muscle thickness to determine the level of muscle activity has no clinical relevance in the hip abductor muscles gluteus medius and minimus.

M-mode measurements of muscle motion indicated two main findings. Firstly, the sequence of activity-related motion in the superficial gluteus medius, the deep gluteus medius and gluteus minimus designate a functional differentiation between muscle levels in different depths of the hip abductors. Secondly and in support of the primary research hypothesis, M-mode greyscale patterns point at a significant difference in the use of gluteus minimus in subjects with non-traumatic, anterior hip pain.

The observation and measurement of activity-related muscle motion by M-mode US may enhance the understanding of the coordination of superficial and deep muscles and provide new insights into muscular contributions to pathology and their therapy.

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Wild JJ (1950) The use of ultrasonic pulses for the measurement of biologic tissues and the detection of tissue density changes. *Surgery* 27: 183-188.

Willcox E, Burden A (2013) The influence of varying hip angle and pelvis position on muscle recruitment patterns of the hip abductor muscles during the clam exercise. *The Journal of Orthopaedic and Sports Physical Therapy* 43: 325-331.

Winter DA (1990) Biomechanics and motor control of human movement. New York: John Wiley & Sons.

Wright AA, Hegedus EJ (2012) Augmented home exercise program for a 37-year-old female with a clinical presentation of femoroacetabular impingement. *Manual Therapy* 17: 358-363.

Xie H-B, Zheng Y-P, Guo J-Y, Chen X, Shi J (2009) Estimation of wrist angle from sonomyography using support vector machine and artificial neural network models. *Medical Engineering & Physics* 31: 384-391.

Yazbek PM, Ovanessiam V, Martin RL, Fukuda TY (2011) Nonsurgical treatment of acetabular labrum tears: a case series. *Journal of Orthopaedic and Sports Physical Therapy* 41: 346-353.

Young A, Stokes M, Crowe M (1984) Size and strength of the quadriceps muscles of old and young women. *European Journal of Clinical Investigation* 14: 282-287.

Young A, Stokes M, Crowe M (1985) The size and strength of the quadriceps muscles of old and young men. *Clinical Physiology (Oxford, England)* 5: 145-154.

Young CC, Rose SE, Biden EN, Wyatt MP, Sutherland DH (1989) The effect of surface and internal electrodes on the gait of children with cerebral palsy, spastic diplegic type. *Journal of Orthopaedic Research* 7: 732-737.

Yucesoy CA, Huijing PA (2007) Substantial effects of epimuscular myofascial force transmission on muscular mechanics have major implications on spastic muscle and remedial surgery. *Journal of Electromyography and Kinesiology* 17: 664-679.

Zamorani PM, Valle M (2007) Muscle and tendon. In Bianchi G, Martinoli C (Eds) Ultrasound of the Musculoskeletal System. Berlin: Springer.

Zimny NJ (1998) Clinical reasoning in the evaluation and management of undiagnosed chronic hip pain in a young adult. *Physical Therapy* 78: 62-73.

11 Appendices

Appendix 2.1 Publications Rehabilitative Ultrasound Imaging

PUBLICATIONS in PubMed, search 1: 2/09/2011, search 2: 12/07/2012. Search algorithm: "ultrasound imaging" muscle (activ* OR onset OR contraction)

No	Title	Journal	Body region	Year	Measurement
1	[Ultrasound imaging of muscles in Duchenne muscular dystrophy]. Rinsho Shinkeigaku. 1989 Jan;29(1):49-53. Aizawa H, Kozima S, Takagi A.	Rinsho Shink	thigh&shank	1989	echogenicity
2	Ultrasound imaging of the quadriceps muscle in elderly athletes and untrained men. Muscle Nerve. 1991 Jun;14(6):527-33. Sipilä S, Suominen H.	MuscleNerve	thigh	1991	CSA&thickness
3	Masseter muscle thickness measured by ultrasonography and its relation to facial morphology. J Dent Res. 1991 Sep;70(9):1262-5. Kiliaridis S, Kälebo P.	DentalRes	masseter	1991	
4	Skeletal muscle contraction in healthy volunteers: assessment with Doppler tissue imaging. Radiology. 1995 Mar;194(3):837-42. Grubb NR, Fleming A, Sutherland GR, Fox KA.	Radiol	general	1995	motion
5	The influence of posture on perineal ultrasound imaging parameters. Int Urogynecol J Pelvic Floor Dysfunct. 2001;12(2):104-6. Dietz HP, Clarke B.		urol	2001	
6	[Masseter thickness measured by ultrasonography of 50 young healthy adults in relation to facial morphology]. Zhongguo Yi Xue Ke Xue Yuan Xue Bao. 2001 Feb;23(1):60-2. Zhao JZ, Dai Q, Lai QS.	Zhongguo Yi Xue Ke Xue	masseter	2001	
7	M-mode ultrasound: a reliable measure of transversus abdominis thickness? Clin Biomech (Bristol, Avon). 2002 May;17(4):315-7. Bunce SM, Moore AP, Hough AD.	ClinBiomech	trunk	2002	thickness

8	The relation between the transversus abdominis muscles, sacroiliac joint mechanics, and low back pain. Spine (Phila Pa 1976). 2002 Feb 15;27(4):399-405. Richardson CA, Snijders CJ, Hides JA, Damen L, Pas MS, Storm J.	Spine	trunk	2002	motion
9	Measuring mechanical properties of the vastus lateralis tendon-aponeurosis complex in vivo by ultrasound imaging. Scand J Med Sci Sports. 2003 Aug;13(4):259-65. Bojsen-Møller J, Hansen P, Aagaard P, Kjaer M, Magnusson SP.	ScandMedSciSport	thigh	2003	displacement
10	Measurement of muscle contraction with ultrasound imaging. Muscle Nerve. 2003 Jun;27(6):682-92. Hodges PW, Pengel LH, Herbert RD, Gandevia SC.	MuscleNerve	general	2003	Fascicles,thickness
11	Accurate measurement of muscle belly length in the motion analysis laboratory: potential for the assessment of contracture. Gait Posture. 2003 Apr;17(2):119-24. Fry NR, Childs CR, Eve LC, Gough M, Robinson RO, Shortland AP.	GaitPost	shank	2003	length
12	Changes in recruitment of the abdominal muscles in people with low back pain: ultrasound measurement of muscle activity. Spine (Phila Pa 1976). 2004 Nov 15;29(22):2560-6. Ferreira PH, Ferreira ML, Hodges PW.	Spine	trunk	2004	thickness
13	Ultrasound imaging distinguishes between normal and weak muscle. Arch Phys Med Rehabil. 2004 Jun;85(6):980-6. Chi-Fishman G, Hicks JE, Cintas HM, Sonies BC, Gerber LH.	ArchPhysMed	thigh	2004	Width,thickness
14	Paradoxical muscle movement in human standing. J Physiol. 2004 May 1;556(Pt 3):683-9. Loram ID, Maganaris CN, Lakie M.	Jphysiol	shank	2004	displacement
15	Measurement of abdominal muscle thickness using M-mode ultrasound imaging during functional activities. Man Ther. 2004 Feb;9(1):41-4. Bunce SM, Hough AD, Moore AP.	ManTher	trunk	2004	thickness
16	The use of ultrasound imaging of the abdominal drawing-in maneuver in subjects with low back pain. J Orthop Sports Phys Ther. 2005 Jun;35(6):346-55. Teyhen DS, Miltenberger CE, Deiters HM, Del Toro YM, Pulliam JN, Childs JD, Boyles RE, Flynn TW	JOSPT	trunk	2005	thickness
17	Influence of 90-day simulated microgravity on human tendon mechanical properties and the effect of resistive countermeasures. J Appl Physiol. 2005 Jun;98(6):2278-86. Reeves ND, Maganaris CN, Ferretti G, Narici MV.	ApplPhys	shank	2005	stiffness

18	Ultrasound imaging in rehabilitation: just a fad? J Orthop Sports Phys Ther. 2005 Jun;35(6):333-7. Hodges PW.	JOSPT	general	2005	
19	The use of real-time ultrasound imaging for biofeedback of lumbar multifidus muscle contraction in healthy subjects. J Orthop Sports Phys Ther. 2006 Dec;36(12):920-5. Van K, Hides JA, Richardson CA.	JOSPT	trunk	2006	thickness
20	Relationship between low back pain and lumbar multifidus size at different postures. Spine (Phila Pa 1976). 2006 Sep 1;31(19):2258-62. Lee SW, Chan CK, Lam TS, Lam C, Lau NC, Lau RW, Chan ST.	Spine	trunk	2006	CSA
21	Muscle activity onset in the lumbar multifidus muscle recorded simultaneously by ultrasound imaging and intramuscular electromyography. Clin Biomech (Bristol, Avon). 2006 Nov;21(9):905-13. Vasseljen O, Dahl HH, Mork PJ, Torp HG.	ClinBiomech	trunk	2006	deformation
22	An MRI investigation into the function of the transversus abdominis muscle during "drawing-in" of the abdominal wall. Spine (Phila Pa 1976). 2006 Mar 15;31(6):E175-8. Hides J, Wilson S, Stanton W, McMahon S, Keto H, McMahon K, Bryant M, Richardson C.	Spine	trunk	2006	CSA&thickness
23	Resonance in the human medial gastrocnemius muscle during cyclic ankle bending exercise. J Appl Physiol. 2006 Jul;101(1):111-8. Takeshita D, Shibayama A, Muraoka T, Muramatsu T, Nagano A, Fukunaga T, Fukashiro S.	ApplPhys	shank	2006	Fascicles,length
24	The response of the transverse abdominis and internal oblique muscles to different postures. Man Ther. 2006 Feb;11(1):54-60. Ainscough-Potts AM, Morrissey MC, Critchley D.	ManTher	trunk	2006	thickness
25	Ultrasound imaging of the female perineum: the effect of vaginal delivery on pelvic floor dynamics. Ultrasound Obstet Gynecol. 2006 Feb;27(2):183-7. Costantini S, Esposito F, Nadalini C, Lijoi D, Morano S, Lantieri P, Mistrangelo E.		urol	2006	
26	Improved activation of lumbar multifidus following spinal manipulation: a case report applying rehabilitative ultrasound imaging. J Orthop Sports Phys Ther. 2007 Oct;37(10):613-9. Brenner AK, Gill NW, Buscema CJ, Kiesel K.	JOSPT	trunk	2007	thickness

27	A comparison of select trunk muscle thickness change between subjects with low back pain classified in the treatment-based classification system and asymptomatic controls. J Orthop Sports Phys Ther. 2007 Oct;37(10):596-607. Erratum in: J Orthop Sports Phys Ther. 2008 Mar;38(3):161. Matacolla, Carl [corrected to Mattacola, Carl G]. Kiesel KB, Underwood FB, Mattacola CG, Nitz AJ, Malone TR.	JOSPT	trunk	2007	thickness
28	Rehabilitative ultrasound imaging of the posterior paraspinal muscles. J Orthop Sports Phys Ther. 2007 Oct;37(10):581-95. Review. Stokes M, Hides J, Elliott J, Kiesel K, Hodges P.	JOSPT	trunk	2007	thickness
29	Ultrasound imaging assessment of abdominal muscle function during drawing-in of the abdominal wall: an intrarater reliability study. J Orthop Sports Phys Ther. 2007 Aug;37(8):480-6. Hides JA, Miokovic T, Belavý DL, Stanton WR, Richardson CA.	JOSPT	trunk	2007	Length,thickness
30	Assessment of abdominal muscle function during a simulated unilateral weight-bearing task using ultrasound imaging. J Orthop Sports Phys Ther. 2007 Aug;37(8):467-71. Hides JA, Wong I, Wilson SJ, Belavý DL, Richardson CA.	JOSPT	trunk	2007	length&thickness
31	Rationale, design, and protocol for the prevention of LBP in the military (POLM) trial. BMC Musculoskelet Disord. 2007 Sep 14;8:92. George SZ, Childs JD, Teyhen DS, Wu SS, Wright AC, Dugan JL, Robinson ME.	BMCmusk	trunk	2007	thickness
32	Can activity within the external abdominal oblique be measured using real-time ultrasound imaging? Clin Biomech (Bristol, Avon). 2007 Nov;22(9):972-9. John EK, Beith ID.	ClinBiomech	trunk	2007	thickness
33	Fascicle length change of the human tibialis anterior and vastus lateralis during walking. J Orthop Sports Phys Ther. 2007 Jul;37(7):372-9. Chleboun GS, Busic AB, Graham KK, Stuckey HA.	JOSPT	thigh&shank	2007	fascicles
34	Effect of static and ballistic stretching on the muscle-tendon tissue properties. Med Sci Sports Exerc. 2007 Mar;39(3):494-501. Mahieu NN, McNair P, De Muynck M, Stevens V, Blanckaert I, Smits N, Witvrouw E.	MedSciSportEx	shank	2007	stiffness
35	Lack of human muscle architectural adaptation after short-term strength training. Muscle Nerve. 2007 Jan;35(1):78-86. Blazevich AJ, Gill ND, Deans N, Zhou S.	MuscleNerve	thigh	2007	fascicles&thickness
36	Measurement of lumbar multifidus muscle contraction with rehabilitative ultrasound imaging. Man Ther. 2007 May;12(2):161-6. Kiesel K, Uhl T, Underwood F, Rodd D, Nitz A.	ManTher	trunk	2007	thickness

37	Improved contraction of the transversus abdominis following spinal manipulation: a case study using real-time US imaging. Man Ther. 2007 Aug;12(3):280-5. Gill N, Teyhen D, Lee I	ManTher	trunk	2007	thickness
38	Assessment of muscle fatigue using sonomyography: muscle thickness change detected from US images. Med Eng Phys. 2007 May;29(4):472-9. Shi J, Zheng YP, Chen X, Huang QH.	MedEngPhys	upperarm	2007	thickness
39	Three-dimensional ultrasound imaging of the levator hiatus in late pregnancy and associations with delivery outcomes. Aust N Z J Obstet Gynaecol. 2007 Jun;47(3):176-80. Lanzarone V, Dietz HP.		urol	2007	
40	Pelvic floor function in elite nulliparous athletes. Ultrasound Obstet Gynecol. 2007 Jul;30(1):81-5. Kruger JA, Dietz HP, Murphy BA.		urol	2007	
41	Vaginal high-pressure zone assessed by dynamic 3-dimensional ultrasound images of the pelvic floor. Am J Obstet Gynecol. 2007 Jul;197(1):52.e1-7. Jung SA, Pretorius DH, Padda BS, Weinstein MM, Nager CW, den Boer DJ, Mittal RK.		urol	2007	
42	Effects of pelvic floor muscle contraction on anal canal pressure. Am J Physiol Gastrointest Liver Physiol. 2007 Feb;292(2):G565-71. Padda BS, Jung SA, Pretorius D, Nager CW, Den-Boer D, Mittal RK.		urol	2007	
43	Transversus abdominis and obliquus internus activity during pilates exercises: measurement with US scanning. Arch Phys Med Rehabil. 2008 Nov;89(11):2205-12. Endleman I, Critchley DJ.	ArchPhysMed	trunk	2008	thickness
44	Assessment of the mechanical properties of the musculoskeletal system using 2-D and 3-D very high frame rate ultrasound. IEEE Trans Ultrason Ferroelectr Freq Control. 2008 Oct;55(10):2177-90. Deffieux T, Gennisson JL, Tanter M, Fink M.	IEEE	upperarm	2008	displacement
45	A pilot study using Tissue Velocity Ultrasound Imaging (TVI) to assess muscle activity pattern in patients with chronic trapezius myalgia. BMC Musculoskelet Disord. 2008 Sep 24;9:127. Peolsson M, Larsson B, Brodin LA, Gerdle B.	BMCmusk	cervical	2008	deformation
46	An US investigation into the morphology of the human abdominal wall uncovers complex deformation patterns during contraction. Eur J Appl Physiol. 2008 Dec;104(6):1021-30.. Brown SH, McGill SM.	EurApplPhys	trunk	2008	thickness
47	Effect of stabilization training on multifidus muscle cross-sectional area among young elite cricketers with LBP. J Orthop Sports Phys Ther. 2008 Mar;38(3):101-8. Hides JA, Stanton WR, McMahon S, Sims K, Richardson CA.	JOSPT	trunk	2008	CSA

48	The applicability of ultrasound imaging in the assessment of dynamic patella tracking: a preliminary investigation. Knee. 2008 Mar;15(2):125-7. Herrington L, Pearson S.	knee	thigh	2008	displacement
49	Influence of feedback schedule in motor performance and learning of a lumbar multifidus muscle task using rehabilitative ultrasound imaging: a randomized clinical trial. Phys Ther. 2008 Feb;88(2):261-9. Herbert WJ, Heiss DG, Basso DM.	PhyTher	trunk	2008	thickness
50	Persistence of improvements in postural strategies following motor control training in people with recurrent LBP. J Electromyogr Kinesiol. 2008 Aug;18(4):559-67. Tsao H, Hodges PW.	ElectrKin	trunk	2008	thickness
51	Rehabilitative ultrasound measurement of select trunk muscle activation during induced pain. Man Ther. 2008 May;13(2):132-8. Kiesel KB, Uhl T, Underwood FB, Nitz AJ.	ManTher	trunk	2008	thickness
52	Use of transabdominal ultrasound imaging in retraining the pelvic-floor muscles of a woman postpartum. Phys Ther. 2008 Oct;88(10):1208-17..Ariail A, Sears T, Hampton E.	PhyTher	urol	2008	
53	Closure mechanism of the anal canal in women: assessed by three-dimensional ultrasound imaging. Dis Colon Rectum. 2008 Jun;51(6):932-9. Jung SA, Pretorius D, Weinstein M, Nager C, Den-Boer D, Mittal R.		urol	2008	
54	Pelvic floor function in nulliparous women using three-dimensional ultrasound and magnetic resonance imaging. Obstet Gynecol. 2008 Mar;111(3):631-8. Kruger JA, Heap SW, Murphy BA, Dietz HP.		urol	2008	
55	The effect of mode of delivery on pelvic floor functional anatomy. Int Urogynecol J Pelvic Floor Dysfunct. 2008 Mar;19(3):407-16. Toozs-Hobson P, Balmforth J, Cardozo L, Khullar V, Athanasiou S.		urol	2008	
56	A rehabilitative ultrasound imaging investigation of lateral abdominal muscle thickness in healthy aging adults. J Geriatr Phys Ther. 2009;32(2):60-6. Erratum in: J Geriatr Phys Ther. 2009;32(3):110. Stetts DM, Freund JE, Allison SC, Carpenter G.	GeriatrPT	trunk	2009	thickness
57	Reliability of thickness measurements of the dorsal muscles of the upper cervical spine: an ultrasonographic study. J Orthop Sports Phys Ther. 2009 Dec;39(12):850-7. Lin YJ, Chai HM, Wang SF.	JOSPT	cervical	2009	thickness
58	Changes in abdominal muscle thickness measured by ultrasound are not associated with recovery in athletes with longstanding groin pain associated with resisted hip adduction. J Orthop Sports Phys Ther. 2009 Oct;39(10):724-32. Jansen JA, Mens JM, Backx FJ, Stam HJ.	JOSPT	trunk	2009	thickness

59	Rehabilitative ultrasound imaging is a valid measure of trunk muscle size and activation during most isometric sub-maximal contractions: a systematic review. Aust J Physiother. 2009;55(3):153-69. Review. Koppenhaver SL, Hebert JJ, Parent EC, Fritz JM.	AustJPT	trunk	2009	thickness
60	The effect of averaging multiple trials on measurement error during ultrasound imaging of transversus abdominis and lumbar multifidus muscles in individuals with low back pain. J Orthop Sports Phys Ther. 2009 Aug;39(8):604-611. Koppenhaver SL, Parent EC, Teyhen DS, Hebert JJ, Fritz JM.	JOSPT	trunk	2009	thickness
61	Isometric contractions reduce plantar flexor moment, Achilles tendon stiffness, and neuromuscular activity but remove the subsequent effects of stretch. J Appl Physiol. 2009 Oct;107(4):1181-9. Kay AD, Blazevich AJ.	ApplPhys	shank	2009	stiffness
62	Reproducibility of rehabilitative ultrasound imaging for the measurement of abdominal muscle activity: a systematic review. Phys Ther. 2009 Aug;89(8):756-69. Review. Costa LO, Maher CG, Latimer J, Smeets RJ.	PhyTher	trunk	2009	thickness
63	Ultrasound characteristics of the deep abdominal muscles during the active straight leg raise test. Arch Phys Med Rehabil. 2009 May;90(5):761-7. Teyhen DS, Williamson JN, Carlson NH, Suttles ST, O'Laughlin SJ, Whittaker JL, Goffar SL, Childs JD.	ArchPhysMed	trunk	2009	thickness
64	Comprehensive Hand Repetitive Intensive Strengthening Training (CHRIST)-induced morphological changes in muscle size and associated motor improvement in a child with cerebral palsy: an experimenter-blind study. NeuroRehabilitation. 2009;24(2):109-17. Lee DR, You JH, Lee NG, Oh JH, Cha YJ.	Neurorehabil	hand/forearm	2009	CSA
65	Behavior of fascicles and the myotendinous junction of human medial gastrocnemius following eccentric strength training. Muscle Nerve. 2009 Jun;39(6):819-27. Duclay J, Martin A, Duclay A, Cometti G, Pousson M.	MuscleNerve	shank	2009	fascicles
66	The validity of Rehabilitative US Imaging for measurement of trapezius muscle thickness. Man Ther. 2009 Oct;14(5):572-8. O'Sullivan C, Meaney J, Boyle G, Gormley J, Stokes M.	ManTher	cervical	2009	thickness

67	Moderate-duration static stretch reduces active and passive plantar flexor moment but not Achilles tendon stiffness or active muscle length. J Appl Physiol. 2009 Apr;106(4):1249-56. Kay AD, Blazevich AJ.	ApplPhys	shank	2009	thickness
68	Reliability of rehabilitative ultrasound imaging of the transversus abdominis and lumbar multifidus muscles. Arch Phys Med Rehabil. 2009 Jan;90(1):87-94. Koppenhaver SL, Hebert JJ, Fritz JM, Parent EC, Teyhen DS, Magel JS.	ArchPhysMed	trunk	2009	thickness
69	Anatomical predictors of maximum isometric and concentric knee extensor moment. Eur J Appl Physiol. 2009 Apr;105(6):869-78. Blazevich AJ, Coleman DR, Horne S, Cannavan D.	EurApplPhys	thigh	2009	fascicles&thickness
70	The effect of chronic low back pain on size and contraction of the lumbar multifidus muscle. Man Ther. 2009 Oct;14(5):496-500. Wallwork TL, Stanton WR, Freke M, Hides JA.	ManTher	trunk	2009	thickness
71	Altered response of the anterolateral abdominal muscles to simulated weight-bearing in subjects with low back pain. Eur Spine J. 2009 Mar;18(3):410-8. Hides JA, Belavý DL, Cassar L, Williams M, Wilson SJ, Richardson CA.	EurSpine	trunk	2009	length&thickness
72	Measurement of cervical multifidus contraction pattern with ultrasound imaging. J Electromyogr Kinesiol. 2009 Jun;19(3):391-7. Lee JP, Wang CL, Shau YW, Wang SF.	ElectrKin	cervical	2009	thickness
73	Onset in abdominal muscles recorded simultaneously by US imaging and intramuscular EMG. J Electromyogr Kinesiol. 2009 Apr;19(2):e23-31. Vasseljen O, Fladmark AM, Westad C, Torp H.	ElectrKin	trunk	2009	deformation
74	Tomographic ultrasound imaging of the pelvic floor: which levels matter most? Ultrasound Obstet Gynecol. 2009 Jun;33(6):698-703. Dietz HP, Shek KL.		urol	2009	
75	Dynamics of female pelvic floor function using urodynamics, ultrasound and Magnetic Resonance Imaging (MRI). Eur J Obstet Gynecol Reprod Biol. 2009 May;144 Suppl 1:S159-65. Review. Constantinou CE.		urol	2009	
76	Differences in transverse abdominis activation with stable and unstable bridging exercises in individuals with low back pain. N Am J Sports Phys Ther. 2010 Jun;5(2):63-73. Saliba SA, Croy T, Guthrie R, Grooms D, Weltman A, Grindstaff TL.	AMJSPORTSPT	trunk	2010	thickness

77	Reliability and relationship between 2 measurements of transversus abdominis dimension taken during an abdominal drawing-in maneuver using a novel approach of US imaging. J Orthop Sports Phys Ther. 2010 Dec;40(12):826-32. Jhu J, Chai H, Jan M, Wang C, Shau Y, Wang S.	JOSPT	trunk	2010	length&thickness
78	Ultrasonography of the cervical muscles: a critical review of the literature. J Manipulative Physiol Ther. 2010 Oct;33(8):630-7. Review. Javanshir K, Amiri M, Mohseni-Bandpei MA, Rezasoltani A, Fernández-de-las-Peñas C.	ManipPhysiol	cervical	2010	CSA&thickness
79	Muscle architecture predicts maximum strength and is related to activity levels in cerebral palsy. Phys Ther. 2010 Nov;90(11):1619-30. Moreau NG, Simpson KN, Teefey SA, Damiano DL.	PhyTher	thigh	2010	fascicles&thickness&CSA
80	Continuous monitoring of electromyography (EMG), mechanomyography (MMG), sonomyography (SMG) and torque output during ramp and step isometric contractions. Med Eng Phys. 2010 Nov;32(9):1032-42. Guo JY, Zheng YP, Xie HB, Chen X.	MedEngPhys	thigh	2010	width&thickness
81	Assessing contractile ability of the quadriceps muscle using ultrasound imaging. Muscle Nerve. 2010 Oct;42(4):530-8. Delaney S, Worsley P, Warner M, Taylor M, Stokes M.	MuscleNerve	thigh	2010	CSA&thickness&width
82	Real-time visualization of muscle stiffness distribution with ultrasound shear wave imaging during muscle contraction. Muscle Nerve. 2010 Sep;42(3):438-41. Shinohara M, Sabra K, Gennisson JL, Fink M, Tanter M.	MuscleNerve	thigh	2010	stiffness
83	A tissue velocity ultrasound imaging investigation of the dorsal neck muscles during resisted isometric extension. Man Ther. 2010 Dec;15(6):567-73. Peolsson A, Brodin LÅ, Peolsson M.	ManTher	cervical	2010	deformation
84	Effect of physical training on pain sensitivity and trapezius muscle morphology. Muscle Nerve. 2010 Jun;41(6):836-44. Nielsen PK, Andersen LL, Olsen HB, Rosendal L, Sjøgaard G, Søgaard K.	MuscleNerve	cervical	2010	?
85	Modelling human musculoskeletal functional movements using ultrasound imaging. BMC Med Imaging. 2010 May 21;10:9. Peolsson M, Löfstedt T, Vogt S, Stenlund H, Arndt A, Trygg J.	BMCImag	shank	2010	displacement
86	In vivo operational fascicle lengths of vastus lateralis during sub-maximal and maximal cycling. J Biomech. 2010 Aug 26;43(12):2394-9. Austin N, Nilwik R, Herzog W.	JBiomech	thigh	2010	fascicles
87	Changes in transversus abdominis thickness with use of the abdominal drawing-in maneuver during a functional task. PM R. 2010 Mar;2(3):187-94; quiz 226. McGalliard MK, Dedrick GS, Brismée JM, Cook CE, Apte GG, Sizer PS Jr.	PMR	trunk	2010	thickness

88	Location and sequence of muscle onset in deep abdominal muscles measured by different modes of ultrasound imaging. J Electromyogr Kinesiol. 2010 Oct;20(5):994-9. Westad C, Mork PJ, Vasseljen O.	ElectrKin	trunk	2010	displacement& deformation
89	The relationship of transversus abdominis and lumbar multifidus activation and prognostic factors for clinical success with a stabilization exercise program: a cross-sectional study. Arch Phys Med Rehabil. 2010 Jan;91(1):78-85. Hebert JJ, Koppenhaver SL, Magel JS, Fritz JM.	ArchPhysMed	trunk	2010	thickness
90	Concentric muscle contractions before static stretching minimize, but do not remove, stretch-induced force deficits. J Appl Physiol. 2010 Mar;108(3):637-45. Kay AD, Blazevich AJ.	ApplPhys	shank	2010	stiffness
91	Tendinopathy alters mechanical and material properties of the Achilles tendon. J Appl Physiol. 2010 Mar;108(3):670-5. Arya S, Kulig K.	ApplPhys	shank	2010	stiffness
92	A comparison of ultrasound and electromyography measures of force and activation to examine the mechanics of abdominal wall contraction. Clin Biomech (Bristol, Avon). 2010 Feb;25(2):115-23. Brown SH, McGill SM.	ClinBiomech	trunk	2010	thickness
93	Retraining motor control of abdominal muscles among elite cricketers with low back pain. Scand J Med Sci Sports. 2010 Dec;20(6):834-42. Hides JA, Stanton WR, Wilson SJ, Freke M, McMahon S, Sims K.	ScandMed SciSport	trunk	2010	thickness
94	Prevalence of major levator abnormalities in symptomatic patients with an underactive pelvic floor contraction. Int Urogynecol J. 2010 Jul;21(7):861-7. Steensma AB, Konstantinovic ML, Burger CW, de Ridder D, Timmerman D, Deprest J.		urol	2010	
95	Mechanisms of pelvic floor muscle function and the effect on the urethra during a cough. Eur Urol. 2010 Jun;57(6):1101-10. Lovegrove Jones RC, Peng Q, Stokes M, Humphrey VF, Payne C, Constantinou CE.		urol	2010	
96	The movement of the diaphragm monitored by ultrasound imaging: preliminary findings of diaphragm movements in classical singing. Logoped Phoniatr Vocol. 2010 Oct;35(3):105-12. Pettersen V, Eggebø TM.		diaphragm	2010	
97	Investigation of Abdominal Muscle Thickness Changes After Spinal Manipulation in Patients Who Meet a Clinical Prediction Rule for Lumbar Stabilization. J Orthop Sports Phys Ther. 2011 Jul 12. Konitzer LN, Gill NW, Koppenhaver SL.	JOSPT	trunk	2011	thickness

98	Ultrasound measurement of transversus abdominis during loaded, functional tasks in asymptomatic individuals: rater reliability. PM R. 2011 Aug;3(8):697-705. Watson T, McPherson S, Fleeman S.	PMR	trunk	2011	thickness
99	Effect of 2 lumbar spine postures on transversus abdominis muscle thickness during a voluntary contraction in people with and without low back pain. J Manipulative Physiol Ther. 2011 Mar-Apr;34(3):164-72. Pinto RZ, Ferreira PH, Franco MR, Ferreira ML, Ferreira MC, Teixeira-Salmela LF, Maher CG.	ManipulPhysiolTher	trunk	2011	thickness
100	Changes in fascicle lengths and pennation angles do not contribute to residual force enhancement/depression in voluntary contractions. J Appl Biomech. 2011 Feb;27(1):64-73. Tilp M, Steib S, Schappacher-Tilp G, Herzog W.	ApplBiomech	shank	2011	fascicles
101	The response of the abdominal muscles to pelvic floor muscle contraction in women with and without stress urinary incontinence using ultrasound imaging. Neurourol Urodyn. 2011 Jan;30(1):117-20. Arab AM, Chehrehrazi M.		urol	2011	thickness
102	Immediate effects of lumbar spine manipulation on the resting and contraction thickness of transversus abdominis in asymptomatic individuals. J Orthop Sports Phys Ther. 2011 Jan;41(1):13-21. Puentedura EJ, Landers MR, Hurt K, Meissner M, Mills J, Young D.	JOSPT	trunk	2011	thickness
103	Investigation of optimal cues to instruction for pelvic floor muscle contraction: A pilot study using 2D ultrasound imaging in pre-menopausal, nulliparous, continent women. Neurourol Urodyn. 2011 Mar 10. Crotty K, Bartram CI, Pitkin J, Cairns MC, Taylor PC, Dorey G, Chatoor D.		urol	2011	
104	Visualization of pelvic floor reflex and voluntary contractions. Stud Health Technol Inform. 2011;163:138-43. Constantinou CE, Korenblum D, Chen B.		urol	2011	
105	Reproducibility of US measurement of transversus abdominis during loaded, functional tasks in asymptomatic young adults. PM R. 2012 Jun;4(6):402-12. McPherson SL, Watson T.	PMR	trunk	2012	thickness
106	US imaging with speckle tracking of cervical muscle deformation and deformation rate: Isometric contraction of patients after anterior cervical decompression and fusion for cervical disc disease and controls. Man Ther. 2012 Jun 13. Peolsson A, Löfstedt T, Trygg J, Peolsson M.	ManTher	cervical	2012	deformation

107	The effect of traditional bridging or suspension-exercise bridging on lateral abdominal thickness in individuals with low back pain. J Sport Rehabil. 2012 May;21(2):151-60. Guthrie R, Grindstaff T, Croy T, Ingersoll C, Saliba S.	Sport Reha	trunk	2012	thickness
108	Comparison of changes in the contraction of the lateral abdominal muscles between the abdominal drawing-in maneuver and breathe held at the maximum expiratory level. Man Ther. 2012 May 15. Ishida H, Hirose R, Watanabe S.	ManTher	trunk	2012	thickness
109	Association between history and physical examination factors and change in lumbar multifidus muscle thickness after spinal manipulation in patients with low back pain. J Electromyogr Kinesiol. 2012 Apr 17. Koppenhaver SL, Fritz JM, Hebert JJ, Kawchuk GN, Parent EC, Gill NW, Childs JD, Teyhen DS.	ElectrKin	trunk	2012	thickness
110	Rehabilitative ultrasound measurement of trapezius muscle contractile states in people with mild shoulder pain. Man Ther. 2012 Apr;17(2):139-44. O'Sullivan C, McCarthy Persson U, Blake C, Stokes M.	ManTher	cervical	2012	thickness
111	Effect of core stability exercises on feed-forward activation of deep abdominal muscles in chronic low back pain: a randomized controlled trial. Spine (Phila Pa 1976). 2012 Jun 1;37(13):1101-8. Vasseljen O, Unsgaard-Tøndel M, Westad C, Mork P	Spine	trunk	2012	motion
112	Novel augmented ADIM training using ultrasound imaging and electromyography in adults with core instability. J Back Musculoskelet Rehabil. 2011;24(4):233-40. Lee NG, Jung JH, You JS, Kang SK, Lee DR, Kwon OY, Jeon HS.	BackMskReha	trunk	2012	thickness
113	The effect of traditional bridging or suspension-exercise bridging on lateral abdominal thickness in individuals with low back pain. J Sport Rehabil. 2012 May;21(2):151-60. Guthrie R, Grindstaff T, Croy T, Ingersoll C, Saliba S.	SportReha	trunk	2012	thickness
114	Comparison of lateral abdominal muscle thickness between weightlifters and matched controls. Phys Ther Sport. 2011 Nov;12(4):171-4. Sitilertpisan P, Pirunsan U, Puangmali A, Ratanapinunchai J, Kiatwattanacharoen S, Neamin H, Laskin JJ.	PTSport	trunk	2011	thickness

115	Tomographic ultrasound imaging of the pelvic floor in nulliparous pregnant women: limits of normality. Ultrasound Obstet Gynecol. 2012 Jun;39(6):698-703. Adisuroso T, Shek KL, Dietz HP.		urol	2012	
116	Changes in lateral abdominal muscle thickness during an abdominal drawing-in maneuver in individuals with and without low back pain. Res Sports Med. 2011 Oct;19(4):271-82. Beazell JR, Grindstaff TL, Hart JM, Magrum EM, Cullaty M, Shen FH.	ResSpMed	trunk	2012	thickness
117	Functional morphology of anal sphincter complex unveiled by high definition anal manometery and three dimensional ultrasound imaging. Neurogastroenterol Motil. 2011 Nov;23(11):1013-9, e460. Raizada V, Bhargava V, Karsten A, Mittal RK.		urol	2011	
118	Tibialis anterior architecture, strength, and gait in individuals with cerebral palsy. Muscle Nerve. 2011 Oct;44(4):509-17. Bland DC, Prosser LA, Bellini LA, Alter KE, Damiano DL.	MuscleNerve	shank	2011	fascicles
119	Rapid force generation is impaired in cerebral palsy and is related to decreased muscle size and functional mobility. Gait Posture. 2012 Jan;35(1):154-8. Moreau N, Falvo M, Damiano D.	GaitPost	thigh	2012	fascicles
120	Differences in transverse abdominis activation with stable and unstable bridging exercises in individuals with low back pain. N Am J Sports Phys Ther. 2010 Jun;5(2):63-73. Saliba SA, Croy T, Guthrie R, Grooms D, Weltman A, Grindstaff TL.	nAmSPPT	trunk	2010	thickness

Appendix 2.2 Pennation angle and PCSA lower limb

Architectural parameters of lower limb muscles, data from Ward et al (2009)

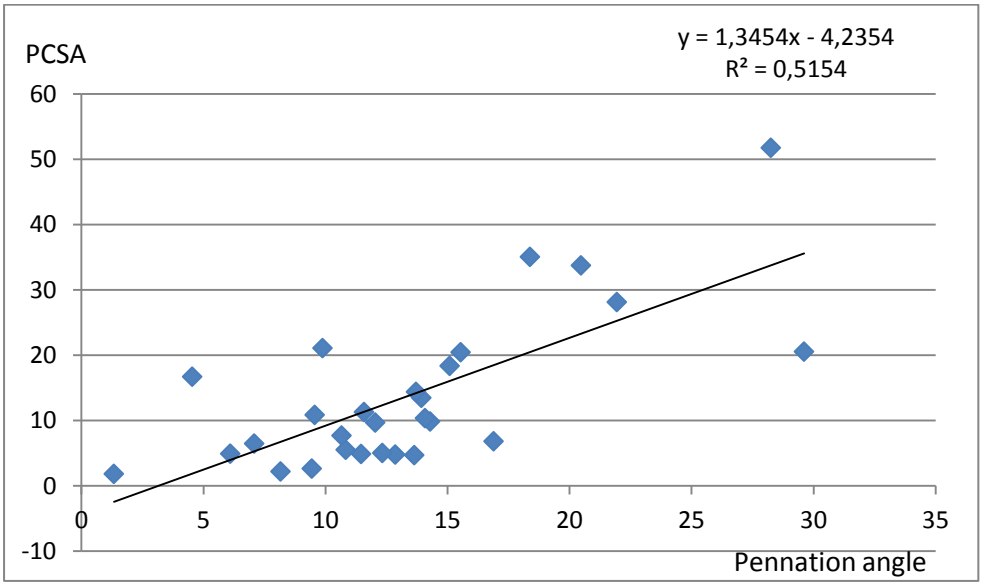
fibre length	pennation angle	PCSA	Muscle
40.3	1,33	1.86	sartorius
22.78	8.16	2.23	gracilis
7.48	9.44	2.67	ext halluc
4.46	13.64	4.73	flex digit
19.3	12.86	4.82	semitendinosus
4.54	11.46	4.91	peroneus br
10.31	6.1	4.95	add br
11.03	12.33	5.06	biceps fem short head
6.93	10.83	5.55	ext digit
10.82	7.08	6.5	add lg
5.27	16.89	6.85	flexor hall
11.69	10.66	7.73	psoas
5.88	12.04	9.72	gastrocn LH
10.66	14.29	9.88	iliacus
5.08	14.08	10.39	peroneus lg
6.83	9.56	10.89	tibialis ant
9.76	11.58	11.33	biceps fem long head
7.59	13.93	13.51	rect fem
3.78	13.71	14.42	tib post
9.93	4.54	16.74	vast intermed
6.9	15.09	18.4	semimembr
14.44	15.54	20.48	add mag
9.68	29.61	20.58	vast med
5.1	9.88	21.12	gastrocn MH
15.69	21.94	28.17	gmax
7.33	20.47	33.78	gmed
9.94	18.38	35.09	vastus later
4.4	28.25	51.79	soleus

Rank of gluteus medius from architectural parameters, in comparison to other lower limb muscles

- **Fibre length**, rank 12 of 28, comparable to ext digit, ext halluc, semimembr, rect fem and tib ant
- **Pennation angle**, rank 25 of 28, comparable to vast lat, flex hall, add mag, gmax soleus and vast med
- **PCSA**, rank 26 of 28, comparable to gmax, gastrocn, vast lat and soleus

Correlation architectural parameters

r= 0.71792554(pennation angle - PCSA)
r= -0.36880652(fibre length - pennation angle)
r= -0.25692724(fibre length - PCSA)



Appendix 3.1 Permission to use Figure 3.4, Primal Pictures Ltd.

use of a part of a hip section image in a publication

Catriona Kerr [catriona@primalpictures.com]

Friday, 1 October 2010 8:17 PM

Dear Angela,

Yes of course you can have permission, could you credit us Image courtesy and copyright Primal Pictures Ltd www.primalpictures.com

Best wishes

Catriona Kerr

Angela Veronika Dieterich

Friday, 1 October 2010 6:00 PM

Thesis

To: pat@primalpictures.com

Hi,

As a part of my research I used a detail from a saggittal hip cadaver section from anatomy.tv for comparison. Now, I want to submit an article to the 'Journal of Orthopaedic and Sports Physical Therapy' including this image and I ask for the permission to do this / the conditions for it.

Kind regards,
Angela

Angela V. Dieterich

PhD candidate| School of Physiotherapy

Curtin University of Technology | GPO Box U1987 | Perth | Western Australia 6845

Telephone +61 8 9266 3667 | Email a.dieterich@postgrad.curtin.edu.au

Appendix 3.2 Interference EMG – US imaging

Date: Tue, 22 Oct 2002 16:36:31 +0800 (WST)
From: Geoff Strauss <G.Strauss@curtin.edu.au>
To: Multiple recipients of list PT_RESEARCH_MUSCULOSKELETAL <pt_resear
Subject: FW: Bortec

FYI - with an expectation that this info will get to [redacted]

Geoffrey R Strauss
Entry Level Programs Coordinator
School of Physiotherapy
Curtin University of Technology

> ----------
> From: [redacted]
> Reply To:
> Sent: Tuesday, October 22, 2002 16:14 PM
> To: Geoff Strauss
> Subject: Bortec
>
> Geoff,
>
> Your emg system is in final testing an should be shipped next week.
>
> Regarding your question about the noise from the UltraSound. Following is
> an extract from their answer:
>
> With regard to the level of noise that could be induced into a system from
> a Doppler ultrasound, there should not be much noise in the EMG as most of
> the emissions from
> the ultra-sound machine should be of a frequency higher than passes
> through
> our low-pass filters (i.e. 1000 Hz). The main reason why the older system
> seems to allow more noise in, is there is one less filtering step than the
> new systems. Also, the new pre-amplifiers are quieter with the new
> systems.
> In order to minimize the noise in the older and newer systems, I would
> first do a max. voluntary contraction (held for a few seconds) or
> stimulate
> a muscle to gain a maximal stimulus response (i.e. M-wave). Do a FFT on
> this and compare it with the FFT from a repeated contraction using the
> same
> parameters as above. Subtracting one spectrum from another should give an
> indication of what frequencies the ultra-sound is producing that might be
> added. Anything within the main power of the EMG signal should be analysed
> by doing repeated MVC's (as above) to average out any differences in power
> that
> might be due to variability in the EMG signal. Off-line filtering using
> your acquisition software, after noise analysis, should clean up the
> signal
> very nicely.
>
> Lets know if this helps or if you require further information.
>
> Regards,
> John.
>
> J C Measurements Pty Ltd
> John Corcoran
> Ph +61 7 5546 8000
> Fax +61 7 5547 0800
>

Appendix 3.3. Subject Information Sheet Setup One

Curtin
University of Technology
School of Physiotherapy

Subject Information Sheet

The use of B-mode, M-mode, and Pulse Waved Doppler ultrasound to determine the timing and magnitude of muscle activation in the gluteus medius and minimus muscles

Thank you for your interest to support this research

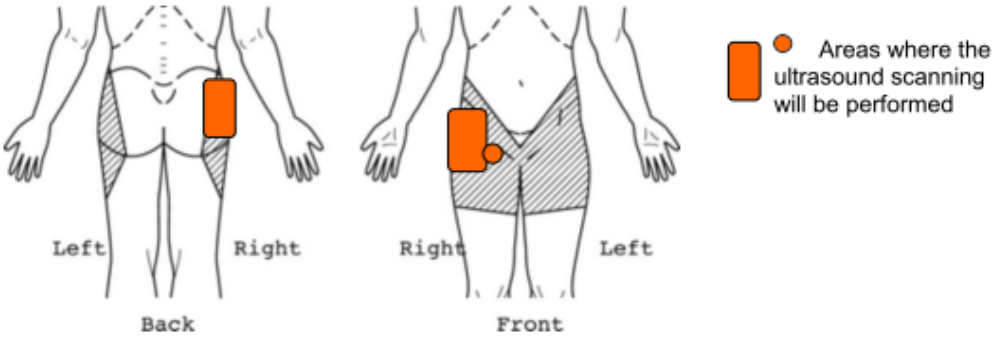
Physiotherapy Investigator: Angela Dieterich…. tel: 9266 3667
Supervisors: Dr Christine Pickard…….. tel: 9266 3643 (School of Physiotherapy)
Mr Geoff Strauss…….. tel: 9266 3689 (School of Physiotherapy)
Ms Louise Deshon…….. tel: 9266 3550 (Discipline of Medical Imaging)

Background Information:

Muscles are organised in deep and superficial layers. It is only partly known how the deep muscles work compared to the superficial. However, there is evidence which hint at an important role of deep muscles in pathology, and also for successful therapy. The currently used modalities to examine the activation of deep muscles are either invasive and painful (fine wire electromyography), or rarely accessible and expensive (Magnetic Resonance Imaging). Ultrasound imaging is supposed to reveal details of deep muscle activation, but with regard to muscle activation the different modes have neither been investigated thoroughly, nor in hip muscles.

This research aims to determine the abilities of different ultrasound modes (B-mode, M-mode, and Pulse-Waved Doppler) to show the muscle activation of two lateral hip muscles, the gluteus medius and minimus muscles. It will be investigated where the muscle activation starts, how both muscles work together to reach different levels of force, and how long they take to relax.

The shaded body region is meant by the term ‚hip region'



(Birrell, F et al, 2005. Defining hip pain for population studies. Ann Rheum Dis, 64: 95-98)

Procedure:

As a participant of the study you will be asked your name, age, body weight, and height. The information you provide is confidential.

You will be requested to undress your lower half except your underwear. You will be offered a bike shorts in which the investigation area is cut out. The outside of your right hip will be palpated, and the hip and the pelvic bone will be outlined with a removable pen.

You will lie on a plinth. Following skin preparation, 3 surface electrodes will be stuck to the skin of your right hip and your pelvis to record the superficial muscle activation. No electric current will be transmitted by these electrodes. The ultrasound head will be spread with gel to ensure good contact with the skin for the ultrasound transmission. The exact probe positions will be tested and marked. Furthermore, the centre of your hip joint will be determined in the groin (●) by use of ultrasound imaging. Then, you will be instructed to push your right leg out to the side (abduction) against resistance. After some repetitions you will be asked to perform the exercise with as much effort as possible. For the strength testing it is important that you try hard to exert your maximal force. This test for maximal strength will be repeated 3 times with breaks in between.

You will take your right leg out to the side against an apparatus, which measures muscle strength. You will see the target force level on a screen display, and also how much force you actually exert. You will be prompted to achieve the target force level (20, 40, 60, and 80 % of your maximal strength) quickly, sustain your force for 4 seconds, and, as quickly as you can, relax your muscles fully. After each exercise you will have a break. Altogether, you will perform 51 repetitions of abduction, which will be recorded by different ultrasound modes or probe positions. The testing is expected to take approximately 1 hour.
You will be asked for one more testing occasion in the days following the first test. On the second occasion the same procedure will be followed. The testing is not complete before the second occasion, because the different measures have to be repeated to ensure their reliability.

The Human Research Ethics Committee of Curtin University has given approval for this study. You are free to withdraw at any time without prejudice. However, the investigator would appreciate prior notice of this intention. All information is confidential and you will not be identifiable in the results or reports from this study. There will be no cost incurred by you for participating in the study. The information gained from the study will be the property of Curtin University of Technology. It will be stored securely in the School of Physiotherapy for five years.

The risks associated with this study are minimal:
- The ultrasound exposure of approximately 25 minutes is within all safety recommendations and recognised as harmless
- Skin preparation for EMG recording involves the recording areas (3 x 9cm^2) being washed with alcohol, abraded and gel applied. This may cause slight discomfort
- Slight delayed onset muscle soreness might occur in exceptional cases. It is expected to subside in 24 – 72 hours
- There is a possibility of irritation of the hip joint's tendons or bursae occurring as a consequence of the repetitive testing, but it is minimal. If you experience pain during the tests, please, inform the investigator immediately. The test will be stopped then.

Your participation in the study will allow physiotherapists to further understand muscle activation and provide a more targeted treatment of muscle dysfunction. Furthermore, you will know how to activate your gluteus medius and minimus muscles, which stabilise your pelvis and hip joint.

Ms Angela Dieterich, Dr Christine Pickard and Mr Geoff Strauss will be happy to answer any queries that you might have about the procedures to be used in the study.

Appendix 3.4 Consent Form Setup One

Curtin
University of Technology
School of Physiotherapy

CONSENT FORM

The use of B-mode, M-mode, and pulse Waved Doppler ultrasound to determine the timing and magnitude of muscle activation in the gluteus medius and minimus muscles

Investigator: Angela Dieterich
Supervisors: Dr. Christine Pickard, Geoff Strauss
Associate Supervisor: Louise Deshon

Declaration of Consent

I, _____
the undersigned *PLEASE PRINT*

of _____

_____

POSTCODE _____ PHONE _____

consent to this investigation. I have read the Information Sheet and understand the consequences and risks associated with participation. I understand the procedure involved in the imaging. I agree to have any results from this study used in any report or research paper, on the understanding that anonymity will be preserved. The investigator has answered my questions concerning the procedures to be used in this study. I may withdraw, either before or during the study, without prejudice. If so, I will contact the investigator (Angela Dieterich, tel: 92663667) at the earliest opportunity. I confirm I am over 18 years old. I am under no duress to participate.

SIGNATURE _____ DATE _____

I have explained the research procedures to which the subject has consented to participate and answered all questions.

RESEARCHER _____ DATE _____

WITNESS _____ DATE _____

Appendix 3.5 Protocol Setup One

Explain, informed consent

↓

data form, random no. in Labview & US

↓

mark pelvis in stance

↓

supine, leg flexed, mark gr. trochanter, the middle of the line, 2 probe and electrode positions

↓

prepare skin, palpate for fem. head

↓

image and mark hip joint centre

↓

electrodes, **skin impedance**

↓

leglift normalisation, note result!

↓

position Mecmesin, give **instructions**, correct it

↓

thigh length, mark foot position, hip extension

↓

position the transducer, adjust B-mode image

↓

start camera, Trigger, remote control

Appendix 3.6 Risk assessment fine-wire electromyography (from application for ethical approval) Setup Two

Appendix B: Potential risks sorted after their likelihood, their severity, their minimisation and their management

Risk	Probability[1]	Severity	Minimisation	Management
Pain 1. Pain during insertion of the fine-wire electrodes 2. Pain induced by the electrodes themselves 3. Discomfort by the abrading procedure in preparation for the surface electrodes	1. High 2. Exceptional 3. Low, was not reported by the subjects in the previous research when they were asked	1. Individually different, ranges from needle prick to sharp but short pain 2. Supposed to induce awareness but not pain 3. Slight discomfort at maximum	1. Insertion perpendicular not parallel to muscle fibre direction is recommended to reduce the pain. Experience of Dr Gibson. Selection of insertion area where the necessary depth to reach the target muscles is low. 2. Use of highly flexible fine-wire (a sample is attached for inspection. Mark that this is the wire within the teflon coat) 3. reduce abrading to the essential minimum, use finest sand paper	1. Volunteers will be fully informed about the range of possible pain. Withdrawal will be possible at any time. 2. Once inserted the electrodes must not be painful. In the exceptional case that the pain lasts the test will be stopped and the electrodes will be removed immediately 3. Subjects will be fully informed. After removal of the surface electrodes they will be offered skin cream.
Vegetative reaction	medium	Individually different, ranges from faint feeling to moments of dizziness. Transient, harmless if falls are prevented.	Insertion in supine on a plinth. Awareness of the subject to avoid fear	Subjects will be lying on a plinth during and following insertion. They will be observed and their legs will be rested in an elevated position if symptoms like paleness occur.
Bruising	low	Blue spot in the tissue, harmless	Assess location of vessels by Colour Doppler ultrasound before insertion. Follow recommendations for electrode positioning	Subjects will be fully informed about that risk, which is completely harmless and does not need further attention. The blue spot is supposed to vanish within maximum 10 days.
Infection	Minimal, this risk accompanies every invasive procedure	Medium, would need medical treatment	Thoroughly sterile procedure	Prepared electrodes and needles will be sterilised. Gloves will be worn. The usual procedures for injections will be followed (disinfection, sterile handling, sharp's bin). In case of painful red swelling following the test session, medical treatment will be provided.
Fine-wire breakage	Minimal, no reports has been found with stainless-steel fine-wire electrodes. Dr Gibson and Dr Rodrigues did not experience any case.	Highly undesired but harmless: a piece of inert metal of the thickness of a thin human hair and most probably the length of 1-3 mm would remain	Choice of stainless steel not silver fine-wire (silver has been reported to break). Choice of fine-wire diameter according to recommendations for safe use (Hodges 2006). When preparing the electrodes, bend the wire only once to prevent predetermined breaking points	Fine-wire electrodes will have a standardised length to recognise a breakage (which may be undetected otherwise). If the fine-wire is protruding it will be carefully removed with a forceps. If the fine-wire is internal, Dr Rodrigues, a neurologist with extensive experience in fine-wire insertion, consented to act as independent advisor on the further procedure.

[1] gauged after reports from experienced colleagues and literature (Hodges 2006)

Appendix B: Potential risks sorted after their likelihood, their severity, their minimisation and their management

As for the risks of the fine-wire procedure we want to raise awareness to the following points:

- Fine-wire EMG is in use since the early 1960s (Basmajian 1962)
- In some neurologic assessments it is a routine clinical instrument with widespread use
- The most remarkable clinical application is for functional neuromuscular stimulation (in patients with paresis), with an implant duration of up to two years using electrode specifications (Daly et al 2001, Marsolais and Kobetic 1986)

Training and experience of Dr Gibson:
Dr Gibson attended a specific training for the insertion of fine-wire electrodes at the University of Aalbourg, Denmark, Center for Sensory Motor Interaction. He performed approximately 60 insertions of fine-wire electrodes in the lower limb. Dr Gibson will attend a second training for the insertion of fine-wire electrodes at the University of Queensland in the beginning of December 2009 to ascertain compliance with Australian standards.

Potential benefits of the study: The study will validate M-mode measurements of muscle activity by the comparison with fine-wire measurements. The aim is to make the fine-wire measurements unnecessary, so that activation patterns of larger samples can be investigated using a method (M-mode ultrasound) with negligible risks.

To whom benefits are likely to accrue: There will be no direct benefit to study participants. Benefits from the results of this research are likely to accrue to hip pain patients, other patient groups who suffer from a pathology in which motor control is a factor of influence (e.g. patients with spasticity) and for physiotherapists. Patients are likely to benefit because currently the activation of their deep muscles can be assessed only with fine-wire EMG. A non-invasive assessment of hip muscle activity will allow for addressing impairments with a wider scope and more individually than currently possible. Physiotherapists are likely to benefit from the study because the investigation will provide a non-invasive tool as well as measurement variables for assessing motor control parameters.

Contingency plan in case of the occurrence of a fine-wire breakage or symptoms alluding to a possible infection (painful redness and swelling in the days following the testing): Dr Rodrigues, a neurologist, has extensive experience with fine-wire insertion and agreed to examine the subject and give advise on the further procedure in case of an adverse event. Dr Rodrigues is a fully qualified clinical neurologist and researcher, has extensive experience with the insertion of intramuscular electrodes and performs this technique routinely in his clinic.

Basmajian J (1962) A new bipolar electrode for electromyography. *Journal of Applied Physiology* 17.

Daly JJ, Kollar K, Debogorski AA, Strasshofer B, Marsolais EB, Scheiner A, Snyder S, Ruff RL (2001) Performance of an intramuscular electrode during functional neuromuscular stimulation for gait training post stroke. *Journal of Rehabilitation Research and Development* 38: 513-526.

Appendix 3.7 Subject information Setup Two

Curtin University of Technology
School of Physiotherapy

Rehabilitative Ultrasound Imaging of deep and superficial hip muscles for assessing motor control in subjects with anterior hip pain

Subject Information **Study 1** 10 October 2009

You are invited to participate in this study, which aims to enable a broader assessment of hip muscle activity than currently possible by using a pain free and harmless method: ultrasound imaging

Why is this study important?
Muscles are organised in deep and superficial layers. Evidence suggests the deep muscles are of specific importance to functional movements. M-mode ultrasound has been found to demonstrate the activation of deep and superficial muscles, but the value of the M-mode measurements needs to be examined by comparisons with the standard method, electromyography (EMG). EMG has the disadvantage that deep muscles can only be assessed with electrodes which have to be inserted into the muscle. The insertion is comparable to an injection into the muscle. This study aims to establish the examination of deep hip muscle activity with M-mode ultrasound, which is pain free and free of risks.

What is involved?
We will examine muscles at the side and back of the hip (gluteus medius, minimus and maximus, quadratus femoris). We will provide specially designed trousers to protect your privacy and enable access to the muscles. You will be asked about your age, body weight and general level of physical activity, and you will draw your subject number. All information you provide is confidential. All further data will be labelled with your subject number, not your name.

Preparation (lateral hip only): Your hip movement and strength will be examined. You will choose by lot which hip will be recorded. The lateral hip will be palpated and scanned by ultrasound to determine the location of your hip muscles and the best position for the ultrasound transducer and the EMG electrodes. The hip, a hip muscle, the pelvic bone, the main vessel and the transducer and electrode positions will be outlined with a removable pen. Little reflective balls which will indicate your movement will be stuck to the leg and the trunk. Following skin preparation, five surface electrodes will be stuck to the skin of your hip and your abdomen to record the superficial muscle activation. No electric current will be transmitted by these electrodes. The ultrasound transducer will be positioned in a foam block which will be fixed with straps to your pelvis. The inserting area for the fine-wire electrodes will be sterilised. Dr Will Gibson, who is trained and experienced in inserting fine-wire electrodes, will insert the electrodes with a needle which is commonly used for injections. Two insertions for two deep muscle levels are necessary. The leads of all electrodes will be taped to the skin. You will then be asked to push your leg out to the side to test the electrode and transducer function.

Testing: For every task there will be practice trials until you feel comfortable with the task.

Task 1: Push out to the side (abduction), all repeated three times with breaks in between

a) push-out with as much effort as possible
b) push-out in different levels of strength (minimal, 30% and 60 % of your maximal strength, slowly increasing strength). You will see the target level on a screen display, and also how much force you actually exert
c) Move your leg out and in against resistance

Task 2: Stay relaxed when you are moved, three repetitions of each movement
The investigator will move your leg to the side, will bend the other leg and will also move your upper body. All movements should happen without your activity because we are also examining the ultrasound images of passive movements.

Task 3: Standing in front of a screen and lifting an arm, five repetitions with each arm
A small device which indicates arm movement will be attached to your wrist. The screen will indicate to lift and lower either your left or your right arm to the side.

Task 4: Step down from a step of 25 cm height, five repetitions
You will step onto a measurement plate which is embedded in the floor of the movement laboratory. The recorded leg will be the leading leg.

This study is part of a larger investigation on activation patterns of hip muscles. We would appreciate being able to use your data in study 2 as part of the data on muscle activation of individuals without hip pain (information sheet is attached). For including your data, also ultrasound recordings (but not EMG) of two posterior hip muscles during arm lifts (task 3) and step-down (task 4) will be needed. If you consent, the ultrasound transducer will be moved to the posterior muscles and you will be asked to repeat tasks 3 and 4.

The testing will take around 70 minutes for the lateral muscles and 15 minutes for the posterior muscles.

This research is in accordance with the National Statement on Ethical Conduct in Research Involving Humans. The Human Research Ethics Committee of Curtin University has given approval for this study (Nr.). You are free to withdraw at any time without prejudice. All information is confidential and you will not be identifiable in the results or reports from this study. There will be no cost incurred by you for participating in the study. The information gained from the study will be the property of Curtin University of Technology. It will be stored securely in the School of Physiotherapy for five years. If you wish to complain about the way this study was handled, please contact the Secretary of the Human Research Ethics Committee on 9266 2784 or by email: hrec@curtin.edu.au.

The risks associated with this study are:
Pain during the insertion of the electrodes: The pain may range from a needle prick sensation to a short sharp pain, comparable to pain during an injection. The pain should be stop when the electrodes are in place. In the rare case that the pain lasts the electrodes will be removed and the testing will be stopped.

Vegetative reactions (occasional): may result in a transient faint feeling and low blood pressure directly following the insertion. This reaction is only short and harmless. You will be safely lying on the plinth.

Bruising (exceptional): If a small vessel is pierced (there are no larger vessels in the insertion area) bruising can happen, which would result in a harmless blue spot, which may last until 10 days.

Infection (highly exceptional): As for every injection there is a theoretic risk of infection which will be minimised by the routine sterile procedures. If you experience pain and swelling in the days following the insertion, please, contact us as to receive medical advice and treatment if necessary.

Electrode breakage (highly exceptional): Theoretically, the electrodes can break. There are no reports about such an event. We will ensure assistance and if necessary treatment by an experienced medical doctor.

Ultrasound exposure (negligible): maximal 20 minutes of ultrasound exposure are within all safety recommendations and recognised as harmless.

Discomfort (occasional) by the skin abrasion for the surface EMG electrodes: Skin preparation involves the recording areas (5 x 9cm^2) being washed with alcohol, abraded and gel applied, which is harmless. You will be offered skin cream after the test.

Value of your participation: Your participation will allow the assessment of the activity of deep hip muscles by M-mode ultrasound, avoiding the invasive procedure of intramuscular EMG. This assessment will help to diagnose normal and impaired muscle activation patterns, which will improve the individually optimal therapy of hip pathologies.

Mrs Angela Dieterich, Dr Christine Pickard and Mr Geoff Strauss will be happy to answer any queries that you might have about the procedures to be used in the study.

Physiotherapy Investigator: Angela Dieterich.... tel: 9266 3667 (School of Physiotherapy)
Supervisors: Dr Christine Pickard..... tel: 9266 3643 (School of Physiotherapy)
Mr Geoff Strauss..... tel: 9266 3689 (School of Physiotherapy)

If you agree to participate in this study, please complete the attached consent form and return it to one of the investigators. Thank you very much for your interest.

Appendix 3.8. Consent Setup Two

Curtin University of Technology
School of Physiotherapy

Rehabilitative Ultrasound Imaging of deep and superficial hip muscles for assessing motor control in subjects with anterior hip pain

Consent form **Study 1** 9 October 2009

Physiotherapy Investigator: Angela Dieterich.... tel: 9266 3667 (School of Physiotherapy)
Supervisors: Dr Christine Pickard..... tel: 9266 3643 (School of Physiotherapy)
Mr Geoff Strauss..... tel: 9266 3689 (School of Physiotherapy)

CONSENT for the investigation of the lateral hip muscles with EMG and ultrasound

I understand that:

- I am not required to participate in this study
- I can withdraw at any time and I can withdraw the consent to the use of my data at any time, without giving reasons and without prejudice or negative consequences
- I have been informed and I understand the purpose of the study
- I have the right to ask any questions before and during the study
- I am fully aware of the specific risks in this study which are explained in the information sheet
- I will receive advice and if necessary free medical treatment in case of the occurrence of worrying symptoms following the testing
- Information collected from me will be used for health research and will not be revealed to others
- Information collected from me will be confidential and securely stored in the School of Physiotherapy for five years
- Research results will be published without allowing for identification of participants

I agree to participate in the study as outlined to me:

_____ _____
Signature *please, print name*

_____ _____
Date *Subject ID*

CONSENT for the investigation of the posterior hip muscles by ultrasound

I understand that:

- All points signed above apply also for this part of the study

I agree to participate in the examination of the activation of the posterior hip muscles by ultrasound

_____
Signature

This research is in accordance with the National Statement on Ethical Conduct in Research Involving Humans. The Human Research Ethics Committee of Curtin University has given approval for this study (Nr.). You If you wish to complain about the way this study was handled, please contact the Secretary of the Human Research Ethics Committee on 9266 2784 or by email: hrec@curtin.edu.au.

Appendix 3.9 Protocol Setup Two

Fine-wire EMG

(prepared plinth, spare pillows, razor, sterile gel, alcohol swabs, sandpaper, 5 surface electrodes, fw EMG connectors, sterile gloves, persist plus, fw electrodes, OPsite, sterile gaze

- Connect EMG and subject unit, use charged battery box
- Bortec on, switch off channels 5-8, gain 500 TFL, 200 fw, select **AD fw** configuration
- Select graph display of the EMG channels in Nexus, check 0-level, scale
- Connect Mecmesin, synch pin input, Laptop AD, open force feedback
- Music on, prepare plinth and cushions, reading material, belt with foam on the plinth

- Welcome subject, **check consent form** and questionnaire, **age, height, weight**
- Mark subject anatomy of the gluteus medius muscle and TFL, mark insertion with bandaid, **US probe position longitudinal and transverse in still frames**
- Skin preparation for the surface electrodes, attach and connect TFL surface electrodes + ground electrode on channel 1, test sEMG function, adjust gain
- Give sterile glove, sterilize insertion area, cover US probe?, put on sterile gel
- Insert fw electrodes, secure wires, attach amplifiers, cover fw with gaze and OPsite
- Test fw function, adjust gain
- Fix US probe, remove pillows for supine position, retest EMG

- **Passive movement** 1.left leg flex, 2. right leg abd, subject relaxed
- Position leg ortheses, adjust Mecmesin, connect & zero Mecmesin with leg contact
- Show correct movement, practise it, correct it
- **Isometric ramp activation**, abduction for 100% force level Mecmesin, gain ok?
- **Minimal, 30 slow & 60% MVIC slow and rapid,** 3 repetitions after practise
- **Probe artefacts** on 30% activated muscle, 1. Tilt, 2. Rock 3. Glide
- Take off Mecmesin
- **Stance**, adjust transducer, test EMG
- Demonstrate and practise **lateral arm lifts** in stance, 8 repetitions
- Take off US, attach sEMG electrodes gluteus medius
- **Lateral arm lifts**, 6 repetitions
- Back on plinth in side-lying
- Bring leg in extended neutral position and ask subject to hold it for **normalisation**
- Pillows for insertion position
-
- Relaxation break, remove all electrodes, tape insertion spot
- Give evaluation sheet

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May 13, 2012

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Appendix 7.1 Full results of Study Five

Table 11.1. Latencies of M-mode high-frequency pattern and torque onsets relative to EMG onset, averaged per force level and velocity condition; +latency, onset followed EMG onset; -latency, onset preceded EMG onset.

Mean (SD) SEM, ±CI, *MDC*; in ms	Asymptomatic group, n = 9			
	SD800	SD1200	SD2000	F
Gmed to EMG	26 (49) 34, ±17, *96*	51 (69) 49, ±25, *135*	83 (148) 105, ±66, *290*	72 (107) 76, ±38, *210*
Gmin to EMG	0 (64) 45, ±22, *126*	51 (110) 78, ±38, *216*	62 (105) 74, ±43, *205*	102 (91) 64, ±31, *178*
Gmed fast to EMG	10 (16) 11, ±11, *31*	16 (19) 13, ±13, *37*	103 (220) 156, ±153, *432*	48 (26) 19, ±18, *52*
Gmin fast to EMG	11 (28) 20 ±18, *55*	75 (144) 102 ±94, *282*	38 (102) 72 ±71, *200*	53 (40) 28, ±26, *79*

Table 11.2. Latencies of high-frequency pattern onset relative to torque onset, averaged per force level and velocity condition; +latency, onset follows torque onset; -latency, onset precedes torque onset.

Mean (SD, SEM, ±CI) MDC; in ms	Asymptomatic group, n = 9		
	SD800	SD1200	SD2000
Gmed to torque	-54 (126) 89, ±27, *248*	-28 (132) 93, ±31, *259*	15 (203) 144, ±63, *398*
Gmin to torque	-96 (138) 97, ±29, *270*	-35 (157) 11, ±36, *308*	-17 (65) 46, ±19, *128*
Gmed fast to torque	-41 (25) 17, ±16, *48*	-32 (20) 14, ±13, *40*	59 (190) 135, ±124, *373*
Gmin fast to torque	-43 (38) 27 ±25, *75*	21 (123) 87 ±80, *241*	-14 (25) 18 ±18, *50*

Table 11.3. Determination of excitation onset from motion onset: limits of agreement (LOA), correlation (*r*) and prediction (r^2), single trials; n trials, number of included trials.

Mean difference, LOA, in ms, (n trials) *r*, r^2	Asymptomatic group, n = 9		
	SD800	SD1200	SD2000
Gmed to EMG	21, 90 (70)	46, 140 (62)	91, 349 (36)
	0.989, 0.979	0.974, 0.948	0.577, 0.333
Gmin to EMG	0, 184 (84)	53, 267 (72)	43, 165 (45)
	0.948, 0.898	0.895, 0.801	0.968, 0.937

Table 11.4. Determination of electrical excitation by HF M-mode patterns: percentage of trials in which rapid muscle motion preceded EMG onset by more than 10 ms, in brackets percentage of the 'early motion' trials in which earlier motion could be explained by motion in the adjacent muscle, in italics mean time of advance of the unexplained trials.

	Asymptomatic group		
	SD800	SD1200	SD2000
Gmed	12.9% (33.3%) *43 ms*	8.1% (60%) *12 ms*	0.0% (n.a.)
Gmin	38.1% (21.9%) *63 ms*	22.2% (25%) *63 ms*	15.6% (14.3%) *38 ms*

Table 11.5. Subgroup comparison, latencies of EMG and M-mode high-frequency onsets relative to torque onset for each of the four subjects per subgroup; +latency, onset follows torque onset; -latency, onset precedes torque onset.

Mean (SD), 95% C.I.	Symptomatic subjects, n = 4				Asymptomatic subjects, n = 4			
	EMG	SD800	SD1200	SD2000	EMG	SD800	SD1200	SD2000
1 Deep	-91 (45)	-62 (21)	-41 (11)	-17 (27)	2 (49)	13 (102)	36 (90)	-30
Gmed	-135 - -46	-83 - -41	-51 - -30	-44 - 9	-46 - 50	-87 - 112	-52 - 125	
2 Deep	52 (111)	-34 (18)	-25 (18)	-10 (12)	-57 (69) -	-34 (47)	-48 (49)	-47 (85)
Gmed	-57 - 161	-52 - -16	-43 - -7	-21 - 2	124 - 11	-80 - 12	-95 - 0	-131 - 36
3 Deep	224 (237)	-200 (63)	-206 (60)	-123 (77)	76	-33 (38)	-78 (31)	-4 (35)
Gmed	--105 - 553	-261 - -138	-264 - -147	-210 - -35		-69 - 4	-108 - -48	-44 - 36
4 Deep	116 (439)	162 (341) -	670		-37 (24) -	-5 (55)	49 (76)	56 (74)
Gmed	-493 - 724	310 - 635			58 - -15	-53 - 43	-17 - 116	-17 - 129
1 Gmin	1 (22)	-51 (36)	-27 (30)		-70 (32)	-42 (34)	36 (90)	-22
	-20 - 23	-86 - -16	-57 - 2		-101 - -38	-75 - -9	-52 - 125	
2 Gmin	-34 (32)	-37 (15)	-29 (12)	-22 (26)	-42 (47)	-56 (50)	-48 (49)	-57 (94)
	-65 - -3	-52 - -22	-41 - -18	-48 - 3	-88 - 3	-105 - -8	-95 - 0	-148 - 35
3 Gmin	-171 (98)	-207 (63)	-209 (61)	-164 (104)	-81 (67)	-123 (58)	-78 (31)	-41 (26)
	-267 - -74	-269 - -145	-268 - -150	-266 - -63	-147 - -15	-180 - -66	-108 - -48	-67 - -15
4 Gmin	-45 (75)	-42 (31)	38 (22)		-78 (51)	-39 (99)	46 (70)	57 (84)
	-119 - 28	-73 - -12	13 - 63		-122 - -33	--125 - 48	-16 - 107	-17 - 131

Appendix 7.2. Questionnaire evaluation of the fine-wire EMG experience

Curtin
University of Technology
School of Physiotherapy 15/6/2010

Thank you very much for participating in research including the insertion of fine-wire electrodes. This questionnaire is not part of the research project. Your answers will be valuable for a better understanding of the experience of the insertion and for improving the insertion procedure.

1. Did you feel sufficiently informed before the insertion?

_____

2. What information was missing?

_____

3. Could you relax directly before the insertion?

_____

4. What would have helped you to relax more easily?

_____

5. The insertion has different phases. Please, score your pain experience (VAS 0 – 10) of the four phases:

a) Skin penetration
0_____10

b) Placing the needle in the right spot within the muscle:
0_____10

c) Deep fascia penetration:
0_____10

d) Withdrawing the needle:
0_____10

7. Some people experience sympathetic reactions on needle insertion, such as sweating or a faint feeling. Did you have such experiences? Which symptoms?

_____

8. Were your muscle sore after the insertion? For how long?

_____

9. Did the insertion spots swell, itch or were getting red later?

_____

10. Do you have suggestions for improving fine-wire insertion?

_____

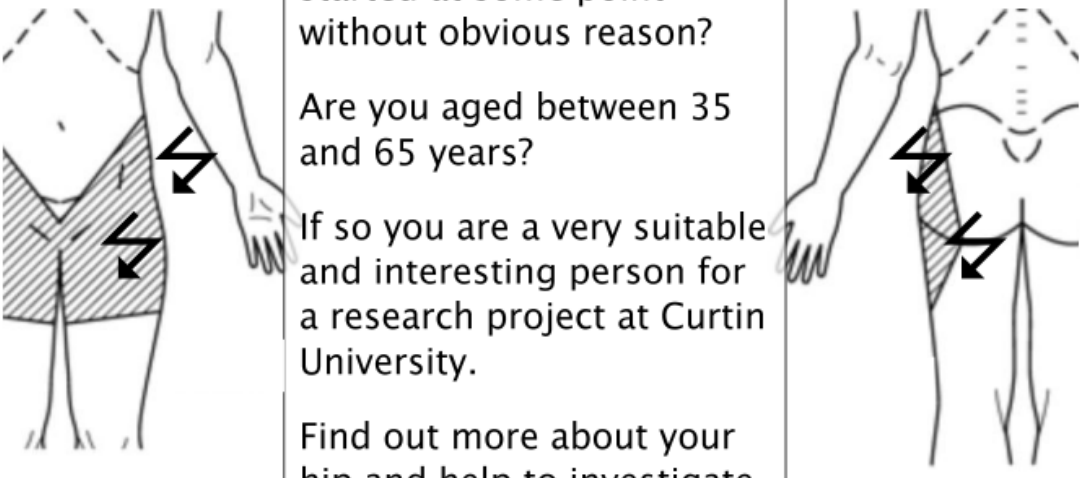
_____

_____ Thanks a lot!

Name: _____ Surname: _____

Appendix 8.1 Call for subjects, Study Six

Do you have **HIP PAIN**?



Do you have hip pain which started at some point without obvious reason?

Are you aged between 35 and 65 years?

If so you are a very suitable and interesting person for a research project at Curtin University.

Find out more about your hip and help to investigate the role of muscles in hip pain.

Participation includes a thorough physical examination of the hip and the recording of muscle activity by ultrasound imaging (appr. 80 min)

Call Angela on 9266 3667,
School of Physiotherapy,
Curtin University of Technology

Angela Dieterich, PhD student at the School of Physiotherapy and the Discipline of Medical Imaging Science, Curtin University of Technology, Kent St, Bentley 6102. Study duration: April – July 2010

Appendix 8.2 Radio Advertising, Study Six

CURTIN FM 100.1 – Free to air message aired for Curtin University of Technology		Contact Details: 9266 3667
SCHOOL/AREA: School of Physiotherapy		
PRODUCT: Hip Pain		**DATE PRODUCED:**
	DURATION: 30 sec	**MUSIC:** **VOICE:**
START DATE: 08/03/10	**END DATE:** 30/06/10	**DAT MASTER:**
JAZ FILE NAME: CHIP		

Version 1

THE SCHOOL OF PHYSIOTHERAPY AT CURTIN, IS SEARCHING FOR PEOPLE WITH HIP PAIN. IF YOU ARE BETWEEN 35 AND 65 YEARS, AND YOU HAVE HIP PAIN WHICH JUST STARTED AT SOME POINT *WITHOUT A CLEAR REASON*, YOU MAY BE ABLE TO HELP WITH IMPORTANT POST-GRADUATE RESEARCH. PLEASE CALL ANGELA AT THE CURTIN UNIVERSITY SCHOOL OF PHYSIOTHERAPY…PHONE 9266 3667, THAT'S 9266 3667.

OR

Version 2

OUCH! IF YOU HAVE HIP PAIN THAT "STARTED"…WITHOUT AN OBVIOUS CAUSE …THEN YOU MAY BE IDEAL FOR A CURTIN UNIVERSITY STUDY.

THE SCHOOL OF PHYSIOTHERAPY NEEDS TO ASSESS PEOPLE BETWEEN 35 AND 65 YEARS OF AGE. SO IF YOUR HIP PAIN JUST "APPEARED' – WITHOUT ANY ACCIDENT– PLEASE CALL ANGELA …A PHD RESEARCHER. IT MAY HELP YOU, AS WELL AS OTHER HIP-PAIN SUFFERRERS.

THAT'S ANGELA AT CURTIN UNIVERSITY…9266 3667…9266 3667

Appendix 8.3 Telephone screening questions, Study Six

1 Telephone Interview _____(date)

Your name:

1. **Can you tell me how old you are?**
2. **Your phone number?**

Can you please tell me about your hip pain? Which side?

When did it start	How did it start	When does it come	How often	How does it feel	Where is it	What do you do with it	Do you know what is wrong with it

Sounds very interesting / Sorry, not the type of hip pain we want to investigate, thank you very much for your interest and your call.

Might be suitable for the study. Let us have a look at the further inclusion criteria.

I have some more questions.

3. Do you or did you have recently any general illness such as rheumatic arthritis or a muscle weakness?

2 Telephone Interview _____(date)

4. Have you had hip surgery?
5. Have you had an accident or surgery of either leg in the past year?
6. Do you have recurrent knee or foot pain?
7. Did you or do you have low back pain at the same time as your hip pain?
8. Do you take medications regularly? Which medications? Against what?
9. Can you please tell me your approximate height and weight

   Height: Weight:
10. Do you sometimes feel dizzy when you have to stand for a while, such as 15 minutes?
11. Can you walk 5 steps without holding the rail?
12. Do you feel that your range of movement in the hip is restricted? In what direction?
13. Can you lift your arms to shoulder level without pain?
14. Do you have x-rays or other imaging of your hip? Can you bring the images when you come?

Thank you very much.

You are very interesting and suitable for this research.

Unfortunately this point is one of the exclusion criteria of the study. Thank you very much for your interest and for calling. It is much appreciated

3 Telephone Interview _____(date)

Would it be possible for you to come here for a physical examination of your hips and ultrasound imaging of your muscles as you perform some simple tasks. It will be lifting an arm in standing, and stepping up and down a step of 20 cm height.

Which days of the week would be possible for you?

Your appointment:

I will send you more information on the study in which you are going to take part, your consent form for the study, and your parking permission with a map.

Your address:

We are going to compare the data of people with hip pain with data from people without hip pain. This is a very critical part of the study to find people without hip pain who fit to the people with hip pain. I would like to ask whether you have a friend or relative without hip pain with whom you would like to come.

I have to talk to them before, to see whether they fit into the study. If you know someone, please, tell her / him to call me so that we can arrange an appointment for you together.

I am looking forward to seeing you on the .....

Thank you very much for calling.

Appendix 8.4 Screening questions control subjects, Study Six

(Telephone screening)

The following questions are for identifying whether you can be included in the study. If you have any questions in between, feel free to ask.

1. What is your age?
2. Do you have any general disease?
   like e.g. rheumatic arthritis or muscle paresis?
3. Have you had a trauma of either leg in the past 6 months?
4. Do you have regularly knee or ankle pain?
5. Do you have any hip issue, like a hip treatment in the past, hip surgery, a hip feeling sometimes dodgy or sensitive after being very active?
6. During the past 12 months, have you had pain in or around either of your hips on most days for a month or longer?
7. Did you or do you have concurrent low back pain with your hip pain?
8. Do you feel dizzy if you have to stand for a while, like 20 minutes?
9. Can you lift either of your legs until horizontal?
10. Can you lift your arms without pain?
11. Can you walk a staircase without holding the rail?
12. Are you currently on any medication regarding pain, cramps or spasms, depression or anxiety?

    If yes, please specify the name of the medicament(s)

    _____

13. Could you please tell me, approximately, how tall you are?_____

    And what is your approximate weight? _____

Thank you very much!

- Make appointment
- Mailing address, contact details
- Organise parking permit
- Send the information sheet, consent form, questionnaires and parking permit latest 10 days before appointment

Appendix 8.5 Cover letter Study Six

FACULTY OF HEALTH SCIENCES
SCHOOL OF PHYSIOTHERAPY

Curtin University of Technology

GPO Box U1987 Perth
Western Australia 6845

TELEPHONE +61 8 9266 4644
FACSIMILE +61 8 9266 3699
EMAIL reception@physio.curtin.edu.au
WEB physiotherapy.curtin.edu.au

26nd of March, 2010

Dear ,

Thank you for your participation in the "Hip Pain Study".
Attached is an information sheet about the study. Please, read it and note if you have any questions. If you feel well informed, please, sign the consent form. If you need to know more we will answer your questions by phone (9266 3667) or at your appointment.

Also attached are two questionnaires.
The WOMAC questionnaire is an internationally recognised assessment of joint pain, joint degeneration and functional limitations. The hip pain questionnaire includes questions about your pain and risk factors for joint problems. Probably, you can answer both questionnaires at home, which would spare you valuable time at your appointment. If you feel uncertain how to answer questions, don't worry, we will do them together.

You will also find a parking voucher for the research parking bay. There is a little map to help you find it. On the appointment day, the parking bay will be blocked by a witches hat with a sign "Hip Pain Study". Please, put the witches hat to the side to use the parking.

Testing will take place in the Movement Analysis Laboratory, which is indicated on the map (pass under the first building, turn right towards a round building with a garage door, the entrance is on the back of this round building).

Your appointment is on_____.

Don't forget your questionnaires and the consent form. If you have x-rays of your hip, please, bring them. They add valuable information to the study about the bony structure of your hip joints, and to the interpretation of your hip pain problem.

We will record hip muscles. To allow good access to these muscles it will be helpful if you wear loose undies (which can be pushed away from the side of the hip) or even, if you have, a G-string. We will provide shorts to wear for the testing.

Looking forward to seeing you,

Angela Dieterich

Appendix 8.6 Subject information Study Six

Curtin University of Technology
School of Physiotherapy

Rehabilitative Ultrasound Imaging of deep and superficial hip muscles for assessing motor control in subjects with non-traumatic hip pain

Subject Information **Study 2** 8 March 2010

You are invited to participate in this study, which uses ultrasound imaging for assessing the muscle activation of people with and without hip pain. This assessment will improve the evidence-based, individual therapy of hip pain pathologies.

Why is this study important?
Muscles are organised in deep and superficial layers. Evidence hints at an important function of the deep muscles. M-mode ultrasound imaging has been found to depict the activation of deep and superficial muscles, but it is unknown whether this delineation is clinically useful. This study will compare the muscle activation of people with and without hip pain. It will also examine the relationship of some clinical symptoms with the muscle activation.

What is involved?
We will examine muscles at the side and back hip (gluteus medius, minimus and maximus, quadratus femoris). We will provide specially designed trousers (cut out on the side) to protect your privacy and enable access to the muscles. You will be asked to fill in three short questionnaires on some personal data, your hip pain and your activities. All information you provide is confidential. You will draw a subject number. All data will be labelled with your subject number, not your name.

Preparation: Your hip mobility and strength will be examined. If you are a hip pain subject, the most painful hip will be recorded. If you are a control subject, we will tell you which hip has to be recorded for comparing the data. The lateral and posterior hip will be palpated and scanned by ultrasound to determine the location of your hip muscles and the best positions for the ultrasound transducer. The hip, a hip muscle, the pelvic bone and the transducer positions will be outlined with a removable pen. Little reflective balls which will indicate your movement will be stuck to the leg and the trunk. The ultrasound transducer will be positioned on a support device which will be fixed with straps to your pelvis. You will then be asked to move the leg to test the transducer function.

Testing: For every task there will be practice trials until you feel comfortable with the task. Both tasks need to be performed once with ultrasound on the lateral hip and once with the ultrasound on the posterior hip.

Task A: Standing in front of a screen and lifting an arm, five repetitions with each arm
A small device which indicates arm movement will be attached to your wrist. The screen will indicate to lift and lower either your left or your right arm to the side or to the front.

Task B: Step down from a step of 20 cm height, five repetitions
You will step onto a measurement plate which is embedded in the floor of the movement laboratory. The recorded leg will be the leading leg.

The testing will take approximately 90 minutes.

Curtin University of Technology
School of Physiotherapy

Rehabilitative Ultrasound Imaging of deep and superficial hip muscles for assessing motor control in subjects with non-traumatic hip pain

Subject Information **Study 2** 8 March 2010

This research is in accordance with the National Statement on Ethical Conduct in Research Involving Humans. This study has bee approved by the Curtin University Human Research Ethics Committee (Approval Number PT 0150). The Committee is comprised of members of the public, academics, lawers, doctors and pastoral carers. Its main role is to protect participants. If needed verification of approval can be obtained either by writing to the Curtin University Human Research Ethics Committee, c/- Office of Research and Development, Curtin University of Technology, GPO Box U1987, Perth, 6845 or by telephoning 9266 2784 or by emailing hrec@curtin.edu.au.
You are free to withdraw from this study at any time without prejudice. All information is confidential and you will not be identifiable in the results or reports from this study. There will be no cost incurred by you for participating in the study. The information gained from the study will be the property of Curtin University of Technology. It will be stored securely in the School of Physiotherapy for five years.

The risks associated with this study are negligible:
- Ultrasound exposure: maximal 15 minutes of ultrasound exposure are within all safety recommendations and recognised as harmless.

- There might be a small risk of delayed onset muscle soreness. If you experience slight muscle pain in the two days following the test, don't be concerned. Delayed onset muscle soreness is completely harmless and will cease after one to three days. It may be relieved by warm showers or baths.

- There might be a small risk of falling in the step-down task. This will be minimised by a rail to use when needed. Also, a researcher will be present all the time.

Value of your participation: Your participation will help to diagnose normal and impaired muscle activation patterns, which will improve the physiotherapy of hip pathologies. You will receive the results of your hip examination, which may be a useful reference for any further hip assessment by a physiotherapist or medical doctor. If something unusual other than muscle activity is observed on the images we will recommend that you visit a doctor to have this followed up. If the results from this study indicate a specific muscular impairment we will inform you.

Mrs Angela Dieterich, Dr Christine Pickard and Mr Geoff Strauss will be happy to answer any queries that you might have about the procedures to be used in the study.

Physiotherapy Investigator: Angela Dieterich…. tel: 9266 3667 (School of Physiotherapy)
Supervisors: Dr Christine Pickard…….. tel: 9266 3643 (School of Physiotherapy)
Mr Geoff Strauss…… tel: 9266 3689 (School of Physiotherapy)

If you agree to participate in this study, please complete the attached consent form and return it to one of the investigators. Thank you very much for your interest.

Appendix 8.7 Consent Study Six

Curtin University of Technology
School of Physiotherapy

Rehabilitative Ultrasound Imaging of deep and superficial hip muscles for assessing motor control in subjects with non-traumatic hip pain

Consent form **Study 2** 9/3/2010

Physiotherapy Investigator: Angela Dieterich…. tel: 9266 3667 (School of Physiotherapy)
Supervisors: Dr Christine Pickard…….. tel: 9266 3643 (School of Physiotherapy)
Mr Geoff Strauss…… tel: 9266 3689 (School of Physiotherapy)

CONSENT

I understand that:

- I am not required to participate in this study
- I can withdraw at any time and I can withdraw the consent to the use of my data at any time, without giving reasons and without prejudice or negative consequences
- I have been informed and I understand the purpose of the study
- I have the right to ask any questions before and during the study
- Information collected from me will be used for health research and will not be revealed to others
- Information collected from me will be confidential and securely stored in the School of Physiotherapy for five years
- Research results will be published without allowing for identification of participants
- I will receive the results of my clinical examination. I will be contacted in the later course of the year if results from this research may be relevant for the treatment of my hip

I agree to participate in the study as outlined to me:

_____ _____
Signature *please, print name*

_____ _____
Date *Subject ID*

This research is in accordance with the National Statement on Ethical Conduct in Research Involving Humans. The Human Research Ethics Committee of Curtin University has given approval for this study (Nr. PT015). You If you wish to complain about the way this study was handled, please contact the Secretary of the Human Research Ethics Committee on 9266 2784 or by email: hrec@curtin.edu.au.

Appendix 8.8 Hip examination Study Six

HIP examination (inclinometer, goniometer, US, gel, clipboard, pen, edge of tibia marked)
All tests graded 1 – 3 (slightly positive, positive, strongly positive)

Observation	Leg length from dorsal	☐ = ☐ ☐
	Leg length from frontal	☐ = ☐ ☐
	Quadriceps atrophy	☐ = ☐ left ☐ right
Demonstrate	Axis of single leg balance: left	☐ **Trendelenburg** ☐ comp ☐ iro ☐ n.a. ☐ ok
Demonstrate	Axis of single leg balance: right	☐ **Trendelenburg** ☐ comp ☐ iro ☐ n.a. ☐ ok
Demonstrate	Squat	☐ < ¼ ☐ pain on ☐ left ☐ right side ☐ ok
Guide	Hip pain by spinal motion?	☐ flex ☐ ext ☐ latfl li ☐ latfl re ☐ ro li ☐ ro re ☐ ok
Supine	Tenderness on palpation left	☐ Gr trochanter ☐ Gmed ☐ TFL ☐ groin ☐ ok
	Tenderness on palpation right	☐ Gr trochanter ☐ Gmed ☐ TFL ☐ groin ☐ ok
Test left	**Faber Test** left	☐ + ☐ ++ _____ ☐ ok
	AROM **flexion** left	☐ < 110° ☐ < 90° ☐ ero/abd ☐ pain ☐ ok
	PROM flexion left	_____° ☐ ero ☐ **Thomas test** ☐ pain ☐ ok
	Resistance flexors left	☐ submaximal ☐ weakness ☐ pain ☐ ok
Test left	**SLR** left	_____° ☐ active ☐ provocation ☐ ok
Test left	**Quadrant Test** left	☐ + ☐ ++ _____ ☐ ok
Test right	**Fabers Test** right	☐ + ☐ ++ _____ ☐ ok
	AROM flexion right	☐ < 110° ☐ < 90° ☐ ero ☐ pain ☐ ok
	PROM flexion right	_____° ☐ ero ☐ **Thomas test** ☐ pain ☐ ok
	Resistance flexors right	☐ other side ☐ weakness ☐ pain ☐ ok
Test right	**SLR** right	_____° ☐ active ☐ provocation ☐ ok
Test right	**Quadrant Test** right	☐ + ☐ ++ _____ ☐ ok
	PROM **ab**duction right	_____° ☐ pain ☐ ok
Test right	**Apprehension** ext ero	☐ + ☐ ++ _____ ☐ ok
	Resistance **ab**duction right	☐ submaximal ☐ weakness ☐ pain ☐ ok
	PROM adduction left	_____° ☐ pain ☐ ok
	Resistance adduction left	☐ submaximal ☐ weakness ☐ pain ☐ ok
Test left	**P**ROM **ab**duction left	_____° ☐ pain ☐ ok
	Apprehension ext ero	☐ + ☐ ++ _____ ☐ ok
	Resistance **ab**duction left	☐ other side ☐ weakness ☐ pain ☐ ok
	PROM adduction right	_____° ☐ pain ☐ ok
	Resistance adduction right	☐ other side ☐ weakness ☐ pain ☐ ok
Sit	**P**ROM **ero** in flexion left	_____° ☐ pain ☐ ok
	Resistance ero in flexion left	☐ submaximal ☐ weakness ☐ pain ☐ ok
	PROM **iro** in flexion left	_____° ☐ pain ☐ ok
	Resistance iro in flexion left	☐ submaximal ☐ weakness ☐ pain ☐ ok
	PROM **ero** in flexion right	_____° ☐ pain ☐ ok
	Resistance ero in flexion right	☐ submaximal ☐ weakness ☐ pain ☐ ok

HIP examination (inclinometer, goniometer, US, gel, clipboard, pen, edge of tibia marked)

All tests graded 1 – 3 (slightly positive, positive, strongly positive)

	PROM **iro** in flexion right	_____° ☐ pain ☐ ok
	Resistance iro in flexion right	☐ submaximal ☐ weakness ☐ pain ☐ ok

Prone	Tenderness on palpation right	☐ pirif ☐ tuber isch ☐ gmax ☐ sulcus SIJ ☐ ok
	Tenderness on palpation left	☐ pirif ☐ tuber isch ☐ gmax ☐ sulcus SIJ ☐ ok
	AROM short extensors right	☐ < 10° ☐ +lordosis ☐ n.a. ☐ pain ☐ ok
	PROM short extensors right	☐ < 10° ☐ groin pain ☐ other pain ☐ ok
	Resistance short extensors right	☐ weakness ☐ groin pain ☐ other pain ☐ ok
	PROM **ero** right	_____° ☐ pain ☐ ok
	PROM **iro** right	_____ ☐ pain ☐ ok
Test	**Craig's** Test anteversion right	_____° ☐ iro ☐ ero
	AROM short extensors left	☐ < 10° ☐ +lordosis ☐ n.a. ☐ pain ☐ ok
	PROM short extensors left	☐ < 10° ☐ groin pain ☐ other pain ☐ ok
	Resistance short extensors left	☐ other side ☐ groin pain ☐ other pain ☐ ok
	PROM **ero** left	_____° ☐ pain ☐ ok
	PROM iro left	_____° ☐ pain ☐ ok
Test	**Craig's** Test anteversion left	_____° ☐ iro ☐ ero

Abbreviations: AROM – active range of movement,comp – compensatory lateral flexion of the trunk to ensure pelvic stability in spite of weak abductors, ero – external rotation , Gmax – gluteus maximus, Gmed – gluteus medius, iro – internal rotation, n.a. – not applicable, pirif – piriformis, PROM – passive range of movement, SIJ – sacroiliac joint, SLR – straight leg raise test, TFL – tensor fasciae latae muscle, tuber isch – tuber ischiadicum. 1 – 3 – grading: 1= slightly remarkable, 2= obvious, 3= significant

Main issue_____

_____

_____

_____

_____

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_____

_____

Appendix 8.9 Harris Hip Score

Harris Hip Score	Hip ID:
	Study Hip: ☐ Left ☐ Right
	Examination Date (MM/DD/YY): / /
	Subject Initials: |____|____|____|
	Medical Record Number:

Interval: _____

Harris Hip Score

Pain *(check one)*

- ☐ None or ignores it (44)
- ☐ Slight, occasional, no compromise in activities (40)
- ☐ Mild pain, no effect on average activities, rarely moderate pain with unusual activity; may take aspirin (30)
- ☐ Moderate Pain, tolerable but makes concession to pain. Some limitation of ordinary activity or work. May require Occasional pain medication stronger than aspirin (20)
- ☐ Marked pain, serious limitation of activities (10)
- ☐ Totally disabled, crippled, pain in bed, bedridden (0)

Limp

- ☐ None (11)
- ☐ Slight (8)
- ☐ Moderate (5)
- ☐ Severe (0)

Support

- ☐ None (11)
- ☐ Cane for long walks (7)
- ☐ Cane most of time (5)
- ☐ One crutch (3)
- ☐ Two canes (2)
- ☐ Two crutches or not able to walk (0)

Distance Walked

- ☐ Unlimited (11)
- ☐ Six blocks (8)
- ☐ Two or three blocks (5)
- ☐ Indoors only (2)
- ☐ Bed and chair only (0)

Sitting

- ☐ Comfortably in ordinary chair for one hour (5)
- ☐ On a high chair for 30 minutes (3)
- ☐ Unable to sit comfortably in any chair (0)

Enter public transportation

- ☐ Yes (1)
- ☐ No (0)

Stairs

- ☐ Normally without using a railing (4)
- ☐ Normally using a railing (2)
- ☐ In any manner (1)
- ☐ Unable to do stairs (0)

Put on Shoes and Socks

- ☐ With ease (4)
- ☐ With difficulty (2)
- ☐ Unable (0)

Absence of Deformity (All yes = 4; Less than 4 =0)

- Less than 30° fixed flexion contracture ☐ Yes ☐ No
- Less than 10° fixed abduction ☐ Yes ☐ No
- Less than 10° fixed internal rotation in extension ☐ Yes ☐ No
- Limb length discrepancy less than 3.2 cm ☐ Yes ☐ No

Range of Motion (**indicates normal*)

- Flexion (*140°) _____
- Abduction (*40°) _____
- Adduction (*40°) _____
- External Rotation (*40°) _____
- Internal Rotation (*40°) _____

Range of Motion Scale

211° - 300° (5)	61° - 100 (2)
161° - 210° (4)	31° - 60° (1)
101° - 160° (3)	0° - 30° (0)

Range of Motion Score _____

Total Harris Hip Score _____

http://www.boulderorthopedics.com/Portals/294/Skins/BOU/pdfs/Harris%20Hip%20Score.pdf

Appendix 8.10 WOMAC osteoarthritis index

WOMAC OSTEOARTHRITIS INDEX VERSION VA3.1

INSTRUCTIONS TO PATIENTS

In Sections A, B, and C questions will be asked in the following format. You should give your answers by putting an " ✗ " on the horizontal line.

EXAMPLES:

1. If you put your " ✗ " at the left of the line as shown below, then you are indicating that you have **no** pain.

2. If you put your " ✗ " at the right end of the line as shown below, then you are indicating that your pain is **extreme**.

3. Please note:
   a) that the further to the right you place your " ✗ " the **more** pain you are experiencing.
   b) that the further to the left you place your " ✗ " the **less** pain you are experiencing.
   c) **please do not** place your " ✗ " **past the end of the line.**

You will be asked to indicate on this type of scale the amount of pain, stiffness or disability you have experienced in the last 48 hours.

Think about your _____ (study joint) when answering the questionnaire. Indicate the severity of your pain, stiffness and physical disability that you feel is caused by arthritis in your _____ (study joint).

Your study joint has been identified for you by your health care professional. If you are unsure which joint is your study joint, please ask before completing the questionnaire.

Australia - V4

Section A

PAIN

Think about the pain you felt in your _____ (study joint) due to your arthritis during the last 48 hours.

(Please mark your answers with an " ✗ " on the horizontal line.)

QUESTION: **How much pain do you have?**

		Study Coordinator Use Only
1. Walking on a flat, even surface.	No Pain — Extreme Pain	PAIN1 _____
2. Going up or down stairs.	No Pain — Extreme Pain	PAIN2 _____
3. At night while in bed, i.e., pain that disturbs your sleep.	No Pain — Extreme Pain	PAIN3 _____
4. Sitting or lying awake in bed.	No Pain — Extreme Pain	PAIN4 _____
5. Standing upright (but not moving).	No Pain — Extreme Pain	PAIN5 _____

Australia - V4

Section B

STIFFNESS

Think about the stiffness (not pain) you felt in your_____ (study joint) due to your arthritis during the last 48 hours.

Stiffness is a sensation of **decreased** ease in moving your joint.

(Please mark your answers with an " ✗ " on the horizontal line.)

	Study Coordinator Use Only
6. How **severe** is your stiffness **after first awakening** in the morning? No Stiffness ——— Extreme Stiffness	STIFF6 ____
7. How **severe** is your stiffness immediately after sitting, lying or resting **later in the day**? No Stiffness ——— Extreme Stiffness	STIFF7 ____

Australia - V4

Section C

DIFFICULTY PERFORMING DAILY ACTIVITIES

Think about the difficulty you had in doing the following daily physical activities due to arthritis in your _____ (study joint) during the last 48 hours. By this we mean **your ability to move around and to look after yourself.** (Please mark your answers with an " ✗ " on the horizontal line.)

QUESTION: **What degree of difficulty do you have?**	Study Coordinator Use Only
8. Going down stairs. No Difficulty — Extreme Difficulty	PFTN8 _____
9. Going up stairs. No Difficulty — Extreme Difficulty	PFTN9 _____
10. Standing up after sitting. No Difficulty — Extreme Difficulty	PFTN10 _____
11. Standing (in one position). No Difficulty — Extreme Difficulty	PFTN11 _____
12. Bending to the floor, i.e., to pick something up. No Difficulty — Extreme Difficulty	PFTN12 _____
13. Walking on a flat, even surface. No Difficulty — Extreme Difficulty	PFTN13 _____

Australia - V4

DIFFICULTY PERFORMING DAILY ACTIVITIES

Think about the difficulty you had in doing the following daily physical activities due to arthritis in your _____ (study joint) during the last 48 hours. By this we mean **your ability to move around and to look after yourself**. (Please mark your answers with an " ✗ " on the horizontal line.)

QUESTION: **What degree of difficulty do you have?**

		Study Coordinator Use Only
14. Getting in or out of a car, or getting on or off a bus.	No Difficulty — Extreme Difficulty	PFTN14 _____
15. Going shopping.	No Difficulty — Extreme Difficulty	PFTN15 _____
16. Putting on your socks or stockings.	No Difficulty — Extreme Difficulty	PFTN16 _____
17. Getting out of bed.	No Difficulty — Extreme Difficulty	PFTN17 _____
18. Taking off your socks or stockings.	No Difficulty — Extreme Difficulty	PFTN18 _____
19. Lying and turning in bed.	No Difficulty — Extreme Difficulty	PFTN19 _____

Australia - V4

DIFFICULTY PERFORMING DAILY ACTIVITIES

Think about the difficulty you had in doing the following daily physical activities due to arthritis in your _____ (study joint) during the last 48 hours. By this we mean **your ability to move around and to look after yourself.** (Please mark your answers with an " ✗ " on the horizontal line.)

QUESTION: **What degree of difficulty do you have?**

		Study Coordinator Use Only
20. Getting in or out of the bath.	No Difficulty — Extreme Difficulty	PFTN20 _____
21. Sitting.	No Difficulty — Extreme Difficulty	PFTN21 _____
22. Getting on or off the toilet.	No Difficulty — Extreme Difficulty	PFTN22 _____
23. Performing heavy domestic duties.	No Difficulty — Extreme Difficulty	PFTN23 _____
24. Performing light domestic duties.	No Difficulty — Extreme Difficulty	PFTN24 _____

Australia - V4

Appendix 8.11 Permission to use the WOMAC osteoarthritis index

WOMAC 3.1®/WOMBAT 3.0©/AUSCAN 3.0© and OGI 8. 0© INDICES
ACADEMIC USER AGREEMENT (2009)

The following agreement defines the conditions under which the WOMAC® 3.0/3.1, WOMBAT© 3.0, AUSCAN© 3.0 and OGI© 8.0 Indices (including their original, alternate language computerised and special feature versions) are provided for use. The conditions are as follows:

1. Use of the Indices is limited to authorised users and their clinical and research associates only.

2. The Indices may not be provided to unauthorised individuals or agencies without prior notification of the originator (Prof Nicholas Bellamy).

3. All copies of the Indices made for research or clinical purposes must bear the Originator's (Prof Nicholas Bellamy) copyright insignia.

4. Commercialisation and resale of any of the Indices is strictly prohibited.

5. Although the use and publication of data collected on the Indices is not limited in any way, the physical form of the Indices must not be published or otherwise displayed in any publication, on the Internet or any other public access medium.

6. Permission for use is non-exclusive.

7. Only alternate-language forms created under the Originator's (Prof Nicholas Bellamy) copyright will be used.

8. User agreements will be confirmed in advance, on a protocol by protocol basis.

9. Use outside the agreed protocol is not permitted.

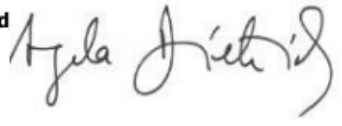
10. The Indices will not be used in any commercial applications, that is, in activities receiving support directly or indirectly, in whole or in part, from any commercial entity (including but not limited to medical and surgical device manufacturers, pharmaceutical companies and biotechnology companies), or in activities in which Index data are provided back to manufacturers with respect to the performance of their products and/or services.

11. The Index will not be modified in any way, or used to create modifications or alternate forms, and will not be used in the development or validation of new outcome measures.

12. The Principle Investigator will provide an original copy of the latest version of the Index User Guide to each person supervising the administration of the Index.

I accept all of the aforementioned conditions to use the WOMAC® VA 3.1 English for Australia Index in a study entitled "Rehabilitative ultrasound imaging of deep and superficial hip muscles for assessing motor control in subjects with anterior hip pain"

in 35 patients at a single point in time

Signed 

Name__Angela Dieterich_____ Date__24/09/2009_____

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Appendix 8.12 Protocol Study Six

Protocol Study 2 "Hip Pain study"	24/3/10

- Witches hat with 'Hip Pain Study' sign in park bay
- 'HIP PAIN STUDY' sign to the two doors of the Motion lab
- Turn on or restart computer and Vicon, laptop, 3 screens. Light 75%
- Connect US and accelerometer, turn US on
- Plug in: synch, remote, channel 1 synch, accelerometer, prepare synch for US
- Prepare 20 markers, 1 wandmarker, tape, washable pen
- Aim cameras, calibrate, set origin of kinematic system
- Prepare belts and transducer supports, research trousers, water, sweeties
- Set-up the instruction screen, stick footprints to the floor and the step, position emergency rail, have a chair behind the US and in the changing cabin
- Prepare phone interview, questionnaires, Harris
- Open new subject and session in Data Management
- Check US PIMS settings, subject screen

- Welcome, introduce, thank, sit down
- Well informed? Questions? – signed consent, random number on all papers
- Questionnaires? Open questions? Harris Hip Score
- Ask for x-rays, have signed consent to copy x-rays, give receipt
- Measure height and weight
- Change into research trouser – meanwhile set up subject in system
- Enter new subject attached to PIG-AD and session, enter US subject
- Mark ASIS, PSIS, PEL markers, gr trochanter, lat epicondyle, tibia edge
- Measure ASIS, knee and ankle width, leg length

- Examination in stance and on the plinth, results with charcoal paper, US for Craigs test, give and write short summary **GIVE A SEAT**
- Position US & accelerometer, footswitch, open subject calibration, control optional markers
- Tick remote, prestart recording 1 sec, control synch setting

- Attach kinematic markers without wand
- Static trial for calibration 1
- Attach wand marker and static calibration 2

- **Need to go anywhere???** Attach accelerometer, plug in synch
- Check US image, attach probe
- Start instruction screen, 2 Armlift task preparation trials
- 10 armlift trials, detach accelerometer
- Step up – step down, preparation trials. Do you feel comfortable with it? 3 each side
- 3 trials left up, 3 trials right up, step-down + walk
- Swap US transducer to extensors, test, refine image, fix probe and belt
- 3 trials left up, 3 trials right up, step-down + walk
- Re-attach accelerometer, 10 armlift trials, detach accelerometer
- Detach US, sit down have a break
- Change
- Thank you very much! Address for x-rays. Results of examination. Put witch hat back.

Reconstruct all trials and export csv. Save on external HD.
Cut and annotate single frames in Xario, burn all on CD.
Lock HD and CDs in office.

File transfer US to PC
Run pipelines and re-label
Finish pipelines and export
Save on external HD

Batch processing: xls files, filters, events & graphs(???)
Reconstruct M-mode?
Measure M-mode frames

Computed pattern detection