Defence Science Journal, Vol 43, No 3, July 1993, pp 201-210

The Effect of External Loads and Cyclic Loading on Normal Patellofemoral Joint Signals

K.O. Ladly[a], C.B. Frank[a,b], G.D. Bell[a,b,c], Y.T. Zhang[a,d] and R.M. Rangayyan[b,d,*]

Sports Medicine Centre[a], McCaig Centre for Joint Injury and Arthritis Research[b], Calgary General Hospital Department of Surgery[c], Department of Electrical and Computer Engineering[d], The University of Calgary, Calgary, Alberta, T2N 1N4, Canada

ABSTRACT

Pain over the anterior portion of the knee joint is a common clinical complaint. A condition known as 'chondromalacia patella' (softening of the cartilage under the patella), which frequently causes anterior knee pain is difficult to diagnose and monitor. Vibrations detected by a contact transducer over the patellofemoral joint may be useful in the assessment of chondromalacia patella. This paper utilised this technique known as vibroarthrography (VAG), to study two potential sources of variability of the normal patellofemoral joint signal. The effect of increased muscular force on the VAG signal was measured by externally loading the joint. The effect of load history (cyclic loading) on the VAG signal was determined by comparing signals before, during, and after application of weights under similar cyclic loading conditions. Results indicated that external loading of the patellofemoral joint caused only minor signal variation. Cyclical loading of the joint, on the other hand, was determined to be a major source of variability of the normal patellofemoral joint signal, which must be controlled in future VAG tests.

1. INTRODUCTION

The patellofemoral (PF) joint is the articulation between the patella (kneecap) and the femur at the front of the knee. Anterior knee pain is a very common clinical condition in which pain occurs around the patella during activities such as ascending and descending the stairs. Pain related to the anterior aspect of the knee has been attributed to many disorders including patellar mechanics[1], disorders of the tissues that support the patella[2] and abnormalities of the cartilage on the articulating surface of the patella[3], called chondromalacia patella. Although normal articular cartilage characteristically deforms under load[4], softened articular cartilage deforms abnormally and begins to breakdown[5]. In the advanced stage, this breakdown progresses to the point where it can cause symptoms of pain and swelling, and may exhibit some associated sounds from the knee. Sounds emitted from the knee joint, in particular from the PF joint, may therefore indicate softening or breakdown of the articular cartilage.

Joint sounds were measured as early as 1885, when a stethoscope was used to detect loose bodies in the human knee joint[6]. This paper considers a more sophisticated Vibroarthrography (VAG) technique for analysis of vibrations (sound) emitted at the joint surfaces. The primary objective in this study was to identify and test two of the potential sources of variation in normal PF joint signals: increased muscle force, and cyclic loading (load history).

2. BACKGROUND

The sounds emitted from joints have long been thought to have diagnostic potential[6-19]. In 1885, a modified stethoscope was used to localise loose bodies in the human knee joint[6]. Auscultation, using a specially designed stethoscope with a rubber diaphragm to prevent skin friction noise, allowed improved sound

---

Received 2 February 1993

[*] To whom all the correspondence is to be addressed

detection[7]. A researcher in 1929 recorded joint sounds by auscultation from 1600 joints and concluded that sounds could be detected before symptoms of early disease were apparent[18]. In 1933, the first electronic recording of joint sounds was performed with a contact microphone, and the signals were graphically recorded on paper[9]. The introduction of microphone recording reduced the subjectivity of knee sound analysis[10,20]. Although the introduction of the computer improved the statistical analysis of the knee sound signals, microphone systems were limited by signal distortion caused by background noise and their poor response to low-frequency signals[8,13,21]. To minimise the extraneous noise from microphone systems, investigators measured the signal in an anechoic chamber[8,14-16]. Recently, a group of researchers used a condensor microphone system to record knee sound signals in an anechoic chamber and reported that signals from osteoarthritic knees increased the sound pressure level significantly during weight bearing as compared to signals when the subjects were non-weight bearing. This increase in sound pressure level was also seen, although less significantly, in the normal knee joint signal[16].

Mollan *et al* and his colleagues initiated research in joint vibration signal analysis using accelerometer transducers[14,23]. This group studied artefact problems associated with their measurement system, including the transducers. There are, however, additional sources of variation to normal signals that were not considered by Mollan's group; for instance, the effect of muscular force and the repeated swinging motion of the lower leg during joint vibration testing. This subject requires further study, and forms the focus of the present paper.

3. PREVIOUS WORK ON VAG SIGNAL ANALYSIS

Over the past several years, we have developed a number of signal processing techniques and have begun to validate them along with our signal collection protocols. Several observations have been made which are reported in the following sections.

3.1 Identifying Muscle Contraction Interference

Since muscle contraction is necessary for active knee movement, we noted that this is a potential artefact in all VAG recordings. In order to identify and characterise what we termed muscle contraction interference (MCI), experiments were designed to isolate muscle movement from joint movement by performing isometric and isotonic contractions[24,25]. Our findings suggested that monitoring and reduction of MCI is necessary in order for VAG to be clinically useful.

3.2 Reducing MCI by Adaptive Filtering

We studied various methods to reduce MCI[24, 25]. Special attention was given to two important issues in MCI reduction: (i) detecting and characterising the MCI components for use as the reference signal in adaptive filtering, and (ii) optimising the step size of the non-stationary adaptive filter. On using the adaptive filter, the correlated MCI was estimated and subtracted successfully from knee VAG signals.

3.3 Mathematical Modelling

A mathematical model has been developed[27] to provide theoretical proof for experimental observations by a random process approach for the patellofemoral pulse (PFP) train produced by slow knee movement. This study provided information on how the spectrum is affected by the inter-pulse interval variation of PFP trains, and gave an insight into the relationship between measurable parameters and physiologically relevant parameters.

3.4 Segmentation and Classification

We have performed signal segmentation in two ways concurrently: (i) fixed segmentation[28], and (ii) adaptive segmentation[17]. The first reflection coefficient and the VAG signal standard deviation were used as inputs to a neural network (perceptron) classifier[19] in our preliminary study on VAG pattern classification.

4. METHODS FOR CURRENT STUDY

4.1 Study Population

A normal knee joint was defined as one having no history of trauma, disease, pain or swelling. Subjects were required to have both knee joints qualify as normal. It was assumed that in the normal population if both the right and left knee fit the normal knee definition, then the two knees were equal and would therefore have similar PF joint signals. Therefore, the knee to be tested was randomly chosen by flipping a coin, and the same knee was used on each of the three test days. The normal group consisted of five male and five female volunteer subjects from The University of Calgary between the ages of 15 and 35 years (mean age

26 years). This age range was identified in the literature as a period when common PF joint problems occur[29,30]. The results from three of the ten subjects were analysed in detail.

4.2 Clinical Auscultation

Prior to vibration signal recording, the PF joint and the rectus femoris muscle were auscultated with a stethoscope (Littmann Classic II). The qualitative information documented from auscultation was used in the analysis of PF joint vibration signals to verify that sound was emitted directly over the normal PF joint and the rectus femoris muscle. The auscultation findings were recorded as the knee was extended and flexed at a swing rate of four seconds from approximately 130 degrees flexion to 0 degrees extension and back to the original starting position for the total cycle. The same swing rate and range of joint movement were also used in the vibration test. Four types of information was recorded for each subject: (i) location of the sound (mid-patella, lateral and medial femoral condyles, and rectus femoris muscle belly), (ii) angle of the knee joint at which each sound component was heard, (iii) amplitude of the sound (low, medium and high), and (iv) a qualitative description of the sound (grind, click, pop and other).

4.3 Multi-Channel Vibration Signal Recording

A multi-channel vibration analysis system was used in this study for the purpose of recording the normal PF joint vibration signal on one channel and the rectus femoris muscle vibration signal on a second channel. The system facilitated simultaneous measurement of the joint and muscle signals as well as an electrogoniometer signal for angle recording.

Accelerometers were used to measure the vibration (sound) signals from the patella and the rectus femoris muscle. The Dytran (model 3115A) transducer was chosen because of its small size, high sensitivity and wide frequency range (from 5 Hz to 10 kHz).

One accelerometer was placed directly on the mid-patellar skin surface with double-sided adhesive tape to measure signals over the PF joint, and a second transducer was placed on the rectus femoris muscle belly approximately 15 cm from the proximal pole of the patella to measure muscle signals. The PF joint and rectus femoris muscle signals were recorded simultaneously, so that these two signals could be compared.

An electrogoniometer (which produces a voltage corresponding to the joint angle) was used to record the knee joint angle during the swing movement on another channel. The electrogoniometer was attached with rubber straps on the lateral aspect of the knee with the axis of rotation at the knee joint line.

The signals were amplified by Gould Isolation Preamplifiers (model 11-5407-58) and Gould Universal Amplifiers (model 13-4615- 58) with a bandwidth of 10 Hz to 1 kHz. The signals were simultaneously recorded on an HP 3968A FM instrumentation recorder and viewed on two HP 1222A oscilloscopes. To digitise the signals, a Zenith PC 386 computer using a National Instruments Signal Data Acquisition Board (model AT-MIO16) and the Lab Window software were used. The same computer was also used for adaptive filtering and short-time Fourier analysis. The adaptive filtering and short-time Fourier analysis computer programs were designed by our research group[24,26]. Figure 1 is a schematic diagram of the testing equipment used in the experiment.

Each subject extended and flexed the knee joint while the subject was verbally directed to swing his/her leg at the specified rate. Four trials, each consisting of ten active swings, were conducted and the data were collected on the second, third and fourth swings of each trial. Although ten swings were recorded in each trial, only swings of two to four were analysed for the purpose of the present study.

The four trials each subject was directed to perform were: (i) repeated knee extension and flexion without external load; (ii) repeated knee extension and flexion with a three-pound weight attached just proximal to the ankle joint; (iii) repeated knee extension and flexion with a six-pound weight attached at the ankle joint; and (iv) repeated extension and flexion without any external load. Thus there were four trials consisting of two unloaded (first and fourth) and two externally loaded (second and third) conditions.

The objective of the investigation was to examine two specific sources of variability of the normal PF joint signal. These two sources of variability were increased muscle force and cyclic loading history of the joint. External weights were used to alter muscle force and the effect of cyclic loading the joint was tested by comparing

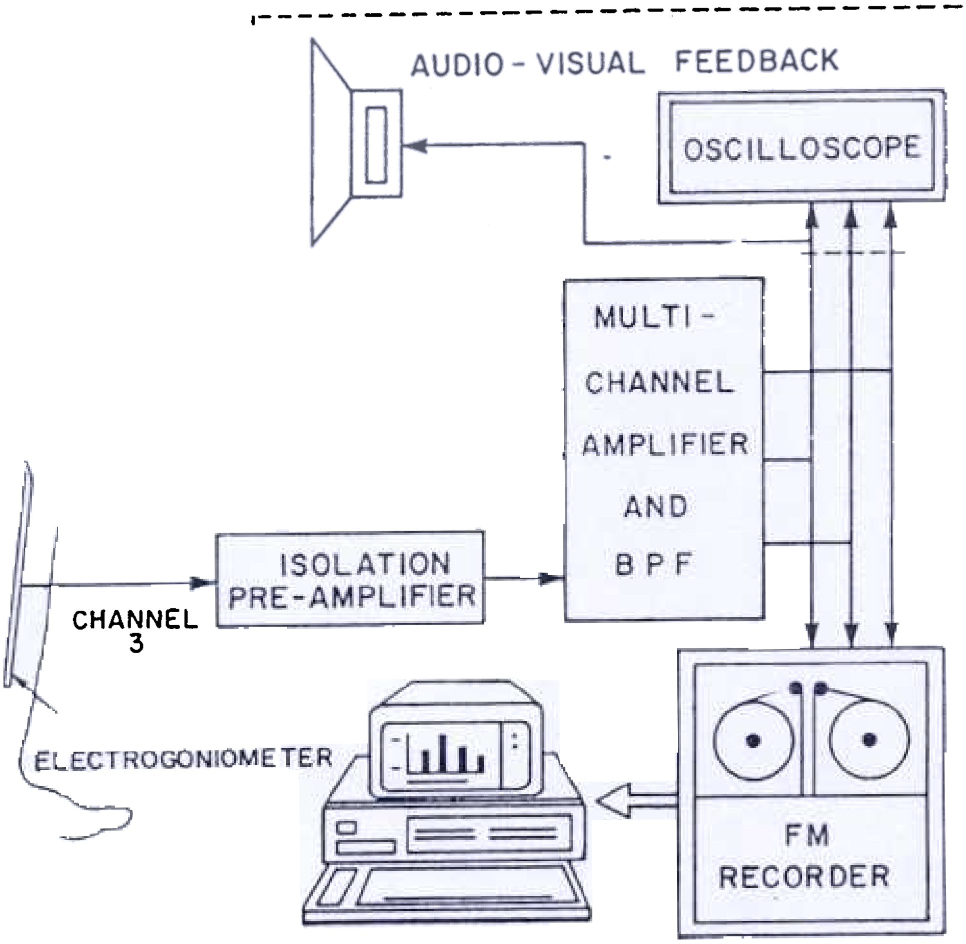


Figure 1. Schematic diagram of the vibration testing equipment and positioning of the transducers and electrogoniometer on the subject

signals across the four trials. The testing protocol is illustrated schematically in Fig. 2.

4.4 Effects of Muscle Force on PF Joint Signals

It was speculated that loading the PF joint would increase its contact force, thereby affecting the resulting PF vibration signal parameters by increasing the contact between the PF joint surfaces. In our preliminary work, a two-dimensional mechanical model was developed to determine theoretically whether external weights of three and six pounds would increase the PF forces during knee extension[31]. The rationale for choosing three- and six-pound weights was that patients with both normal and abnormal knee joints could perform ten consecutive swings with these weights without experiencing pain in the PF joint. Larger weights were not tolerated by patients with abnormal knee joints.

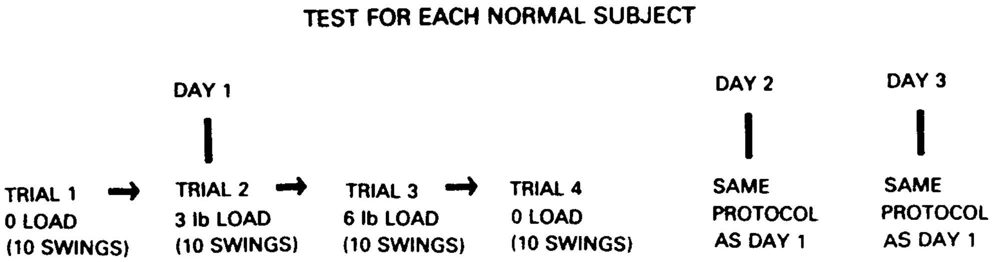


Figure 2. Diagram of testing protocol.

Preliminary results calculated from the mechanical model indicated that maximum PF joint compression force occurred at 40 degrees and that maximum quadriceps force occurred at 90 degrees of knee flexion. The subjects demonstrated a 68 per cent increase in PF compression force when the unloaded test and the three-pound load test were compared. A comparison of the unloaded test and the six-pound load test demonstrated an increase of 119 per cent.

These results indicate that there should be theoretically a considerable increase in the compression effect on the normal PF joint with external loads of three and six pounds on the ankle joint. The estimated angle of maximum PF compression force occurred at 40 degrees. It was therefore predicted that the normal PF joint signal amplitude would increase close to 40 degrees flexion. In addition, if the maximum power is observed at about 40 degrees, it is likely that increased muscle force from the external loads is a source of variance to the signal.

An experiment was then performed to test the effect of increased muscular force on the normal PF joint signal by positioning external loads at the ankle joint. The first and fourth trials had no external load, and the second and third trials were externally loaded by adding weights of three and six pounds. The effect of joint loading was thus measured by comparing the first and second trials, the first and third trials, and the second and third trials.

4.5 Load History

The effect of load history was measured by comparing the fourth swing in the first and fourth trials, which were performed under identical conditions (no weight, same equipment position, same day, same person, same joint, and same joint movement and angle range) to determine whether the loading history during trials two and three would affect the signals measured over the normal PF joint. A measurement of load history (trial one versus four) and load effect was therefore possible for each person and for the entire group.

Subjects were asked not to participate in physical activity on the day prior to the testing session to control day-to-day loading history. In addition, subjects were tested at the same time on each of the three test days to control changes in load history prior to the test.

4.6 Signal Processing

Previous research has established[11,13,17,27] that the knee joint signal is limited to components below 1 kHz. The signals were therefore digitised using a sampling rate of 2 kHz, which provided 8,000 total data points for each swing cycle. A two-step procedure was followed to process the signals: (i) remove the rectus femoris muscle vibration signal using an adaptive filtering process[24,26]; and (ii) perform spectral analysis to determine the range of signal power and median frequency using short-time Fourier analysis techniques.

4.7 Signal Analysis

Each swing cycle was divided into 12 equal segments (20 degrees each) using a fixed segmentation process to divide the signal into specified segment lengths. The rationale for analysing the signal in a fixed segment length was to facilitate comparison of the signal parameters in equal segments.

Normal PF joint signals were compared in the total swing cycle and the last 60 degrees of extension (with the parameters averaged over the last three angle segments). Data from the last 60 degrees also contributed to the total swing analysis. The rationale for choosing the 60 to 0 degree angle range to compare signals is as follows: (i) auscultation results determined that the majority of the sound detected over the patella occurred in the angle range between 60 and 0 degrees of extension, (ii) Kernohan *et al*[12] have suggested that some normal knee joint sounds are emitted in specific angle ranges, and (iii) during the recording of normal PF joint signals, the amplitude was greatest (as viewed on an oscilloscope) between 60 degrees and full knee extension.

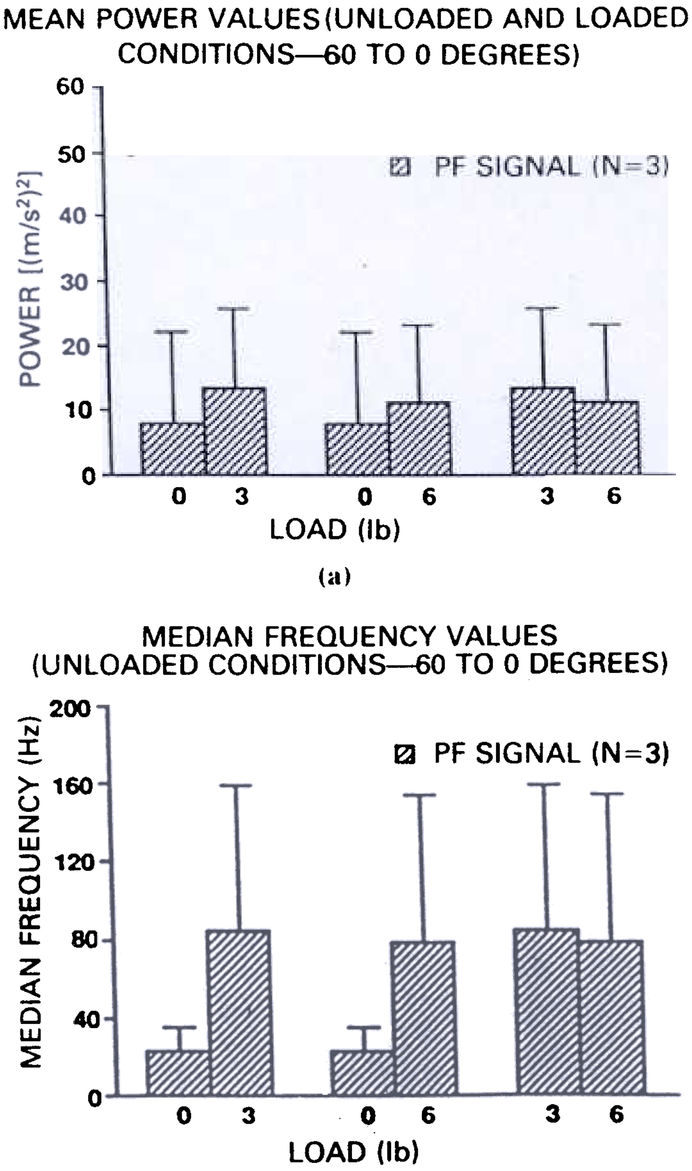
5. RESULTS

Results were analysed for both the total swing and the 60 to 0 degree joint angle range for each signal; however, only the data for the last 60 degrees are presented because the majority of PF joint signals were recorded in this range. The mean value and standard deviation (represented by error bars) are presented graphically for the power and median frequency parameters.

5.1 Effect of External Loads on Normal PF Joint Signals

Results showed only a minimal effect on power with the external loads of three and six pounds over the total swing and the last 60 degrees. A notable increase in median frequency was observed over the total swing with the addition of the three- and six-pound loads. There was a statistically significant increase in the median frequency with the six-pound load (p=0.04). Analysis of the data over the last 60 degrees showed an additional increase in median frequency as a result of the external loads. In this joint angle range some data were missing due to technical equipment failure and thus statistical analysis of the power and median frequency parameters was not possible.

A plot illustrating the effect of external loads on power and median frequency over the last 60 degrees is given in Fig. 3. Data from three subjects and from each of the test pairs analysed over the total swing showed a decrease of 14 per cent in mean power and an increase of 69 per cent in median frequency with the three-pound external load, and an increase of 8 and 75 per cent in mean power and median frequency, respectively, with the six-pound external load. Data analysed over the last 60 degrees showed an increase of the mean power and median frequency by 64 and 271 per cent with the three-pound external load respectively, and 40 and 245 per cent with the six-pound external load. Externally loading the joint also increased the variance in power and median frequency during the last 60 degrees of extension.



(a)

(b)

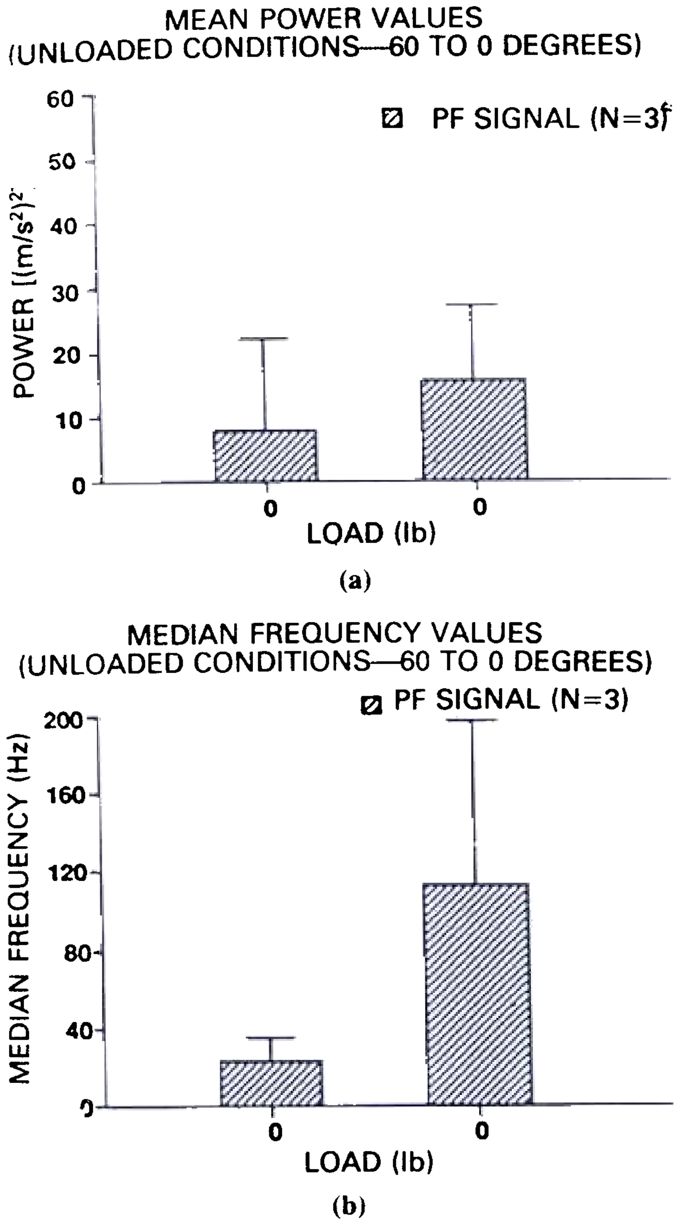
Figure 3. Plots of mean and standard deviation values, (a) power, and (b) median frequency as a function of load of normal PF joint signals for the 60 to 0 degree angle range (3 days).

5.2 Effect of Loading History on Normal PF Joint Signals

Results showed a minimal effect from loading history on mean power over the total swing, and a substantially greater impact on mean power over the last 60 degrees. A comparison of the two unloaded trials one and four showed an increase of 1 per cent in mean power in trial four when the data were averaged over the total swing, and a 90 per cent increase in mean power in trial four when the data were averaged over the final 60 degrees (Fig. 4). There was no statistically significant difference in the power signal parameter between the two unloaded trials when analysed over the total swing (p=0.36).

Median frequency was significantly affected by joint loading history in the total swing. A statistically significant difference was determined in this parameter between the first and fourth unloaded trials (p= 0.006). Furthermore, over the last 60 degrees of extension, the median frequency mean and variance increased (Fig. 4). The median frequency results showed a 398 per cent increase in the mean with the data averaged between 60 and 0 degrees. The statistical significance of these parameters over the last 60 degrees could not be determined due to missing data.

It can be concluded from these results that although power was not statistically different between trials one and four, there was some impact of load history on this parameter. In addition, there was a statistically significant effect of load history on median frequency. Joint loading history therefore was probably a contributing source of variance to normal PF joint signals.



(a)

(b)

Figure 4. Plots of mean and standard deviation values, (a) power, and (b) median frequency as a function of load (trials 1 and 4) of normal PF joint signals for the 60 to 0 degree angle range (3 days).

6. DISCUSSION

6.1 Effect of External Loads on Normal PF Joint Signals

The results indicate that external loads have an insignificant effect on signal power; however, the six-pound load had a statistically significant effect on median frequency. In addition, the external loads increased the variance of both signal parameters in the 60 and 0 degree range, although more dramatically in median frequency.

Two factors probably contributed to the statistically insignificant effect of load on the signal power: (i) the weight may have been insufficient to increase the forces acting on the PF joint to alter the signals; and (ii) the power component of the signal may not be affected by changing external loads.

Researchers have extensively studied the material properties of articular cartilage and have found that cartilage demonstrates a time-dependent deformation response to load[4,5]. Under initial loading there is an instantaneous deformation of the cartilage, which is followed by further time-dependent deformation; once the load is removed, there is a time-dependent recovery phase[4]. The extent of instantaneous deformation was also found to be directly related to the weight of the load[32]. In the present experiment the weight may not have been sufficient to increase the contact force enough to significantly affect cartilage deformation of the patellar or femoral surfaces. Also, friction between surfaces may not have been changed significantly. Minor variation would therefore be expected to occur in the power of a signal being generated by their interaction. If this is correct, power would be a suitable measure of joint vibrations in futute work, as it is not affected by inter-subject differences in muscle force, and knee joint and leg size and weight.

Theoretically, the maximum PF joint force was calculated from the mechanical model at 40 degrees flexion. Consistent with this estimate, the results from this research established that increased power and median frequency occurred in extension between 60 and 0 degrees. However, other reports have indicated that the force transmitted from the patella to the femur may be greater with increased flexion of the knee joint[29,33]. If these latter models are correct, then the signal measured in the last 60 degrees of extension may not be generated from increased articular surface contact.

A statistically significant increase in median frequency was determined in our studies with the six-pound external load. It is possible that median frequency, unlike power, is affected by differences in varying external loads. Therefore, median frequency varies under an externally loaded condition and could also vary between subjects due to their differences in muscle force, knee joint, and leg dimensions.

6.2 Effect of Loading History on Normal PF Joint Signals

The results showed no statistically significant difference in the power between trial one and trial four in the three subjects; however, there was a significant difference in the median frequency between these trials. A further finding was that trial four data showed an increase in mean power, median frequency and variances in the 60 to 0 degree angle range as compared to the total swing. Thus, joint loading history probably affected normal PF joint signals, particularly in the last 60 degrees of movement.

Load history could alter the PF joint surface contact stress. With progressive loading, due to viscoelastic creep, cartilage compresses probably increasing contact area and decreasing contact stress. This could alter signal power or frequency parameters. Further, it seems doubtful that contact stresses would be equal over time unless the patella tracked in exactly the same position with every swing. With a load history, patellar tracking differences probably increase changes in contact stress, and contribute to variability in the recorded signal.

The load history effect on power and median frequency could also have been caused by differences in synovial contact with the PF joint surfaces during the continuous swinging movement. It is possible that the fringe-like folds on the surface of the synovial membrane[34] are in contact with the joint surfaces during the knee extension and flexion swinging motion, and that this contact varies from swing to swing. Furthermore, patellar contact with the synovial membrane is likely increased as the patella moves into the suprapatellar region in the final degrees of knee extension, which is the range in which there is a marked increase in signal variability.

7. CONCLUSION

Based on these results, we conclude that under the conditions described, increased muscular force does not significantly affect normal PF joint signal power, but increases the median frequency. The increase in the mean and variance of the median frequency parameter as a result of the external loads was less than the increase caused by the joint loading history. Thus, variability in individual muscle force, knee joint, and leg dimensions is likely a minor source of variance to the normal PF joint signal. Cyclic loading of the PF joint resulted in an increase in the mean and variance of the signal power and median frequency parameters. The large variances resulting from differences in joint loading histories could potentially cause a large problem in future testing of normal and abnormal patients. By altering the protocol to account for the effect of load history through a standardised exercise just prior to testing, a baseline of exercise could be established for each subject, and this source of variance to the PF joint signal could be controlled considerably. In future tests, in order to minimise variance in comparative experiments, it is essential that load history is controlled.

ACKNOWLEDGEMENTS

This work was supported by grants from the Arthritis Society of Canada, the Nickle Family Foundation of Calgary, and the Olympic Oval Fund. We would like to thank S Tavathia for his contributions to this work.

REFERENCES

1. Insall, J. Chondromalacia patellae: Patellar malalignment syndrome. *Orthopaedic Clinics of North America*, 1979, **10**(1), 117-27.
2. Bray, R.C. & Roth, J.H. Clinical evaluation of anterior knee pain. *Diagnosis*, 1987, **10**, 69-96.
3. Dugdale, T.W. & Barnett, P.R. Historical background: Patellofemoral pain in young people. *Orthopaedic Clin. North Am.*, 1986, **17**(2), 211-19.
4. Albright, J.A. & Brand, R.A. (Eds). Articular cartilage. *In* The scientific basis of orthopaedics. Appleton & Lange, Connecticut, 1987. pp. 347-71.
5. Woo, S.L.Y. & Buckwalter, J.A.(Eds). Articular cartilage. *In* American academy of orthopaedic surgeons symposium on injury and repair of the musculoskeletal soft tissues. American Academy of Orthopaedic Surgeons, Illinois, 1988. pp. 401-82.
6. Heuter, C. (Ed). *Grundriss Der Chirurgie. In Grundriss der chirurgie*, Ed. 3. F.C.W. Vogel, Leipzig, 1885.
7. Blodgett, W.E. Auscultation of the knee joint. *Boston Med. and Surg. J.*, 1902, **146**, 63-66.
8. Chu, M.L.; Gradisar, I.A. & Zavodney, L.D. Possible clinical application of a noninvasive monitoring technique of cartilage damage in pathological knee joints. *J. Clin. Engg.*, 1978, **3**, 19-27.
9. Erb, K.H. *Uber die Moglichkeit der registrierung von*

gelenkgerauschen. *Deutche Ztschr. f. Chir.*, 1933, **241**, 237- 40.

10. Fisher, H. & Johnson, E.W. Analysis of sounds from normal and pathological knee joints. *In* Third International Congress Physical Medicine, 1960. pp. 50-57.
11. Frank, C.B.; Rangayyan, R.M. & Bell, G.D. Analysis of knee joint sound signals for noninvasive diagnosis of cartilage pathology. *IEEE Engg. Med. Biol. Mag.*, 1990, 65-68.
12. Kernohan, W.G.; Beverland, D.E.; McCoy, G.F.; Hamilton, A.; Watson, P. & Mollan, R.A.B. Vibration arthrometry: a preview. *Acta Orthopaedica Scandinavica*, 1990, **61**(1), 70 - 79.
13. McCoy, G.F.; McCrea, J.D.; Beverland, D.E.; Kernohan, W.G. & Mollan, R.A.B. Vibration arthrography as a diagnostic aid in diseases of the knee. *The Journal of Bone and Joint Surgery*, 1987, **69**(2), 288-93.
14. Mollan, R.A.B.; McCullagh, G.C. & Wilson, R.I. A critical appraisal of the auscultation of human joints. *Clinical Orthopaedics and Related Research*, 1982, **170**, 231-37.
15. Nagata, Y. Joint-sounds in gonoarthrosis – clinical application of phonoarthrography for the knees. *Journal Uoeh*, 1988, **10**(1), 47-58.
16. Sasaki, M.; Suzuki, K. & Inoue, J. Analysis and application of joint sounds to osteoarthritis of the knee. *In* Transactions of the Combined Meeting of the Orthopaedic Research Societies of USA., Japan and Canada, 1991. p. 164.
17. Tavathia, S.; Rangayyan, R.M.; Frank, C.B.; Bell, G.D.; Ladly, K.O. & Zhang, Y.T. Analysis of knee vibration signals using linear prediction. *IEEE Trans. Biomed. Engg.*, 1992, **39**(9), 959-70.
18. Walters, C.F. The value of joint auscultation. *The Lancet*, 1929, **1**, 920-21.
19. Zhang, Y.T.; Rangayyan, R.M.; Frank, C.B.; Bell, G.D.; Ladly, K.O. & Liu, Z.Q. Classification of knee sound signals by using neural networks: Preliminary study. *In* Procceedings of the IASTED International Symposium on Expert Systems and Neural Networks, Honolulu, Hawaii, 1990. pp. 61-62.
20. Kernohan, W.G.; Beverland, D.E.; McCoy, G.F.; Shaw, S.N.; Wallace, R.G.H.; McCullagh, G.C. & Mollan, R.A.B. The diagnostic potential of vibration arthrography. *Clin. Orthopaedics and Related Research*, 1986, **210**, 106-12.
21. Chu, M.L.; Gradisar, I.A.; Railey, M.R. & Bowling, G.F. An electro-acoustical technique for the detection of knee joint noise. *Med. Res. Engg.*, 1976, **12**, 18-20.
22. Zhang, Y.T.; Ladly, K.O.; Liu, Z.Q.; Tavathia, S.; Rangayyan, R.M.; Frank, C.B. & Bell, G.D. Interference in displacement vibroarthrography and its adaptive cancellation. *In* Procceedings of the 16th Canadian Medical Biological Engineering Conference, Winnipeg, Canada, 1990. pp. 107-08.
23. Mollan, R.A.B.; Kernohan, G.W. & Watters, P.H. Artefact encountered by the vibration detection system. J. Biomech., 1983, **16**(3), 193-99.
24. Zhang, Y.T.; Frank, C.B.; Rangayyan, R.M.; Bell, G.D. & Ladly, K.O. Step size optimization of nonstationary adaptive filtering for knee sound analysis. *In* Procceedings of the World Congress on Medical Physics and Biomedical Engineering, Kyoto, Japan, 1990. p. 836.
25. Zhang, Y.T.; Ladly, K.O.; Rangayyan, R.M.; Frank, C.B.; Bell, G.D. & Liu, Z.Q. Muscle contraction interference in acceleration vibroarthrography. *In* Procceedings of IEEE/EMBS 12th Annual International Conference, **12**(5), Philadelphia, 1990. pp. 2150-51.
26. Zhang, Y.T.; Rangayyan, R.M.; Frank, C.B. & Bell, G.D. Adaptive cancellation of muscle contraction interference in vibroarthrographic signals. *IEEE Trans. Biomed. Engg.* (Accepted subject to revision).
27. Zhang, Y.T.; Frank, C.B.; Rangayyan, R.M. & Bell, G.D. Mathematical modelling and spectrum analysis of the physiological patellofemoral pulse train produced by slow knee movement. *IEEE Trans. Biomed. Engg.*, 1992, **39**(9), 971- 79.
28. Ladly, K.O.; Frank, C.B.; Bell, G.D.; Zhang, Y.T. & Rangayyan, R.M. The effect of external loads on normal patellofemoral joint vibration signals. *In* Transactions of the Combined Meeting of Orthopaedic Research Societies of USA, Japan and Canada, Banff, Alberta, Canada, 1991. p. 269.
29. Strother, R.T. & Samoil, D. Patellofemoral syndrome: therapeutic regimen based on

biomechanics. Canadian Family Physician, 1989, **35**, 1649-54.

30. Swenson, J.E.; Hough, D.O. & McKeag, D.B. Patellofemoral dysfunction. *Postgraduate Medicine*, 1987, **82**, 125-41.

31. Ladly, K.O. Analysis of patellofemoral joint vibration signals. Department of Medical Science, The University of Calgary, Calgary, Alberta, Canada, 1992. MSc Thesis.

32. Albright, J.A. & Brand, R.A. (Eds). Joint lubrication. *In* The scientific basis of orthopaedics. Appleton & Lange, Connecticut, 1987. pp. 373-86.

33. Huberti, H.H. & Hayes, W.C. Patellofemoral contact pressures. *J. Bone and Joint Surgery* 1984, **66**(5), 715-24.

34. Owen, R.; Goodfellow, J. & Bullough, P. Synovium and synovial fluid. *In* Scientific foundations of orthopaedics and traumatology. W.B. Saunders Company, Philadelphia, 1980. pp. 18-22.