COMPUTER SIMULATIONS TO PREDICT SUBJECT SPECIFIC TISSUE LOADING

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Sporting manoeuvres such as side stepping challenge knee joint stability. The external loads experienced during these tasks do not directly stress ligaments because muscles can also support these loads. Therefore, it is important to have models that can account for how muscles are used. To this end, the development of an electromyography driven knee model is presented. The calibrated model accurately predicts knee flexion-extension moments in a range of static and dynamic tasks. It is then used to examine the contribution of muscles to support varus-valgus loads during static tasks and sporting manoeuvres. In static tasks there are small but specifically directed activation patterns that potentially reduce ligament loading. However, in the sporting manoeuvres the muscular contribution to varus-valgus knee joint stabilisation is much greater. The use of the model in these series of studies shows why it is important to have joint models that account for how people use their muscles. It can then be used to evaluate the efficacy of neuromuscular exercise programmes that may train people to protect knee joint ligaments.

KEY WORDS: EMG, muscle Model, in-vivo muscle loading, knee joint

INTRODUCTION: "Only when the causal relations between applied forces and resultant injury are established and understood can appropriate programs of intervention and prevention be designed and implemented" (Whiting and Zernicke, 1998, p. 177). Moreover, if we are to prevent injury to ligaments, cartilage or bone then the forces experienced by these structures during tasks that cause the insult(s) must be determined. However, determining the actual loads sustained by tissues *in-vivo* is still a problem that is only starting to be resolved.

The methods that we have available to measure *in-vivo* tissues loads are direct measurement or indirect estimation using some form of modelling. Direct measurement is difficult, has obvious serious ethical considerations, and may modify the actual performance of the task. So although modelling does appear a logical choice, there are many problems with this pathway.

First we must consider the factors that have to be accounted for in such models. Let's use the anterior cruciate ligament and the knee joint as an example. The important factors are; the anatomy of the knee, muscle, menisci, and ligaments; the strength of these tissues; the external loading; the static and dynamic joint posture; and the interactions between muscles, ligaments and articular surfaces in the joint. For example, even for the same joint position and load, muscles can be activated quite differently depending the control task (Buchanan & Lloyd, 1995). Loading of the internal structures heavily relies on how muscles are activated, which is person and task specific. Additionally, the indeterminate nature of any joint system must be accounted for.

Electromyograph (EMG) driven joint modelling can take into account all the above factors. EMG driven models have been developed for the lower back (McGill, 1992; Thelen *et al.*, 1994; Nussbaum and Chaffin, 1998), elbow (Buchanan *et al.*, 2000; Soechting and Flanders, 1997), shoulder (Laursen *et al.*, 1998) and knee (White and Winter, 1993; Lloyd and Buchanan, 1996). This paper describes the models developed by my colleagues and I, and summarises how they have been applied to examining subject specific tissue loading at the knee. Two models have been developed, the first a static isometric model, and the second a dynamic model. The models have been used in continuing series of studies to determine the loading of the knee ligaments when the knee is loaded in varus and valgus and internal and external rotation directions. These directions were chosen as the loads produced, when coupled with anterior draw of the tibia, have the potential to highly stress the ligaments, particularly the anterior cruciate ligament of the knee joint (Markolf *et al.*, 1995).

THE MODELS: The EMG driven joint model is a computer simulation in that it predicts the moments produced by the muscles that cross a joint (Besier & Lloyd, 1999; Lloyd et al. 1996, Lloyd & Buchanan, 1996). This model can also be implemented to predict joint motion (see Buchanan et al., 2000).
The model uses "real" data that are typically recorded in a motion analysis laboratory; electromyographs (EMG), 3 dimensional kinematics, and ground reaction force data. These data are recorded from subjects who perform tasks that challenge knee stability. These tasks include static varus-valgus loads, dynamic tasks such as running and cutting, or isometric and isokinetic tasks on a dynamometer.
Both the static and dynamic model consists of 4 parts 1) an anatomical model, 2) an EMG to activation model, 3) a muscle model, and 4) calibration (see Fig. 1). The EMG and motion data recorded during the various tasks are the inputs required by the model to predict joint

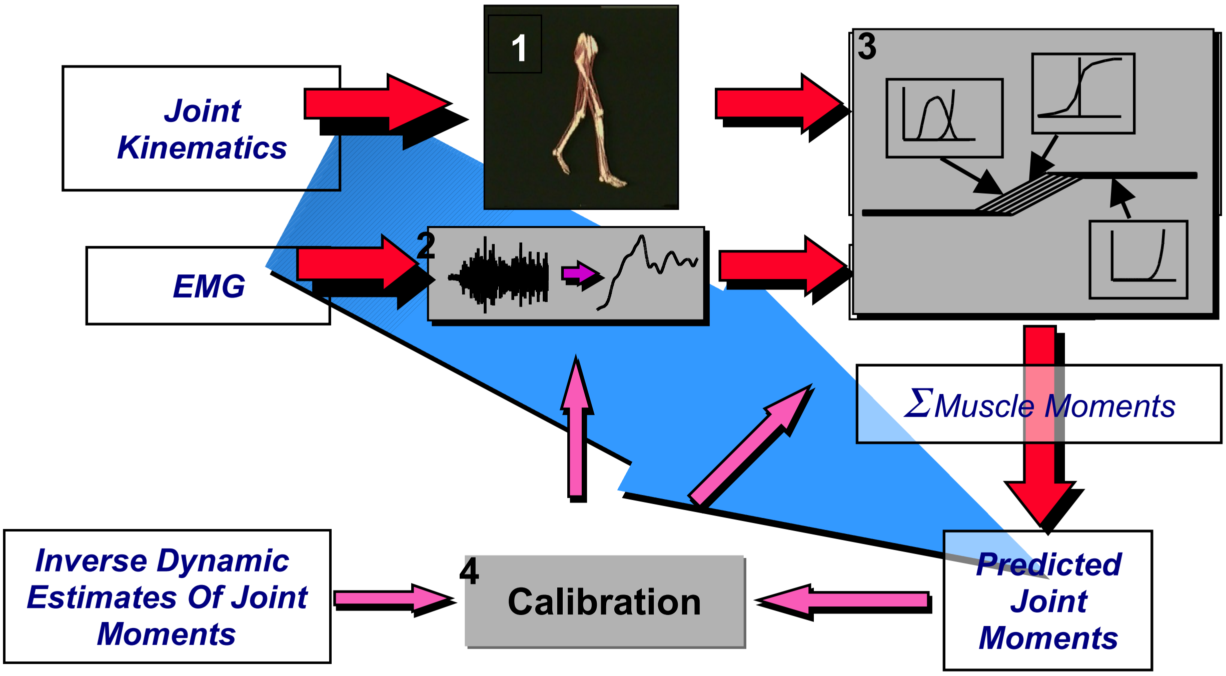

Figure 1 - Schematic of the EMG driven knee joint model showing the sub-components; 1) anatomical model, 2) EMG-to-activation model, 3) muscle model, and 4) calibration.

moments. Joint moments estimated using inverse dynamics are used to first calibrate the model, and then verify the model performance after calibration.
These various sub-components of the models are now discussed.
1. Anatomical model. Software for Interactive Musculoskeletal Modelling (SIMM - Musculographhics) is used to model the anatomy of the lower limbs and knee joint. Using the motion or posture data collected during the trials as input, SIMM estimates the lengths, velocities, and moment arms of the musculotendon units that cross the knee. In the static model, only musculotendon lengths and moment arms are required.
2. EMG to activation model. The output of this model is muscle activation based on the recorded EMG of the each muscle. EMG is first high pass filtered with a 30Hz zero lag Butterworth filter, full wave rectified and then low pass filtered with a 6Hz zero lag Butterworth filter.
In the *static model* the activation is just the average rectified and filtered EMG over the isometric contraction period, with a linear 1-to-1 relationship between processed EMG and activation. In the *dynamic model*, the rectified and filtered EMG is processed using a similar scheme to the linear discrete time dynamic model proposed by Thelen et al. (1994), modified to account for the non-linear EMG to force relationship (Lloyd et al., 1996).

3. Muscle model. Muscle activation, musculotendon lengths and velocities, are used as inputs to determine muscle force employing a Hill-type muscle model similar to that proposed by Zajac (1989). Modifications to this model include a) coupling between activation and optimal fibre length based on the work of Huijing (1996), b) a passive elastic muscle force in the contractile element obtained from an exponential relationship, which allowed for passive forces to be obtained regardless of fibre length (Schutte, 1992), and c) a passive parallel damping element added to the force-velocity relationship as suggested by Schutte (1992) to prevent any singularities of the mass-less model when activation or isometric force are zero.
The net flexion-extension (FE) joint moments generated by the muscles are estimated by multiplying the individual muscle forces with their FE moment arms and summing the subsequent individual muscle FE moments.
4. Calibration. Calibration is performed for each subject using data from a number of different trials. Non-linear optimisation is used to adjust the coefficients in the EMG-to-activation model and muscle model parameters. The optimisation reduces the least-square error between the FE joint moments computed by the model and those estimated by inverse dynamics. Once the model is calibrated and the optimal parameters are obtained, the model is ready to predict individual muscle forces and joint torques.

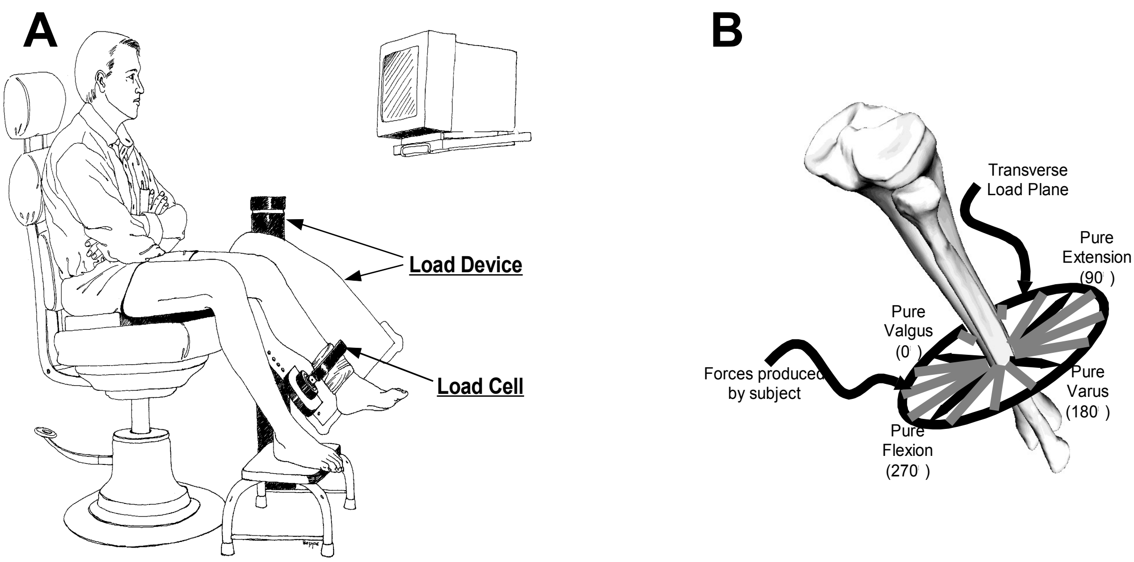

Figure 2 - A) Subject was seated and generated forces in transverse plane at the shin. Subject was given visual display of the target forces and with visual feedback of force they were producing. B) Schematic of the forces produced in the transverse load plane. These forces map into combinations of varus-valgus and flexion-extension moments at the knee. For example, a force produced with a direction between pure extension and pure varus (from 90º to 180º) will map into a moment at the knee that has extension and varus components.

APPLICATION OF THE STATIC MODEL: Muscular support of varus and valgus isometric loads at the human knee. In this study subjects were required to voluntarily generate various forces in a transverse plane just above their ankles, while sitting (Fig. 2A). The forces produced mapped into combinations of varus-valgus and flexion-extension moments at the knee (Fig. 2B). The contributions of their muscles and non-muscular soft tissues (ligaments and joint capsule) to the support of the total external knee joint moment were determined by analysing the experimental data using the EMG driven model of the knee.
Specific model details. The model used in this study is the static version of the model presented above (Lloyd and Buchanan, 1996). The model was calibrated to all trails, with the calibration parameters being muscle flexor and extensor strengths, muscle optimal fibre lengths, and tendon slack lengths.

Muscular contribution to the external knee moments. The fundamental premise in the model is that the muscles contribute 100% of the FE knee load. However, the muscular contribution to the external varus-valgus (VV) knee load has to be calculated. The total VV moment generated by muscles is determined by multiplying the individual muscle forces with

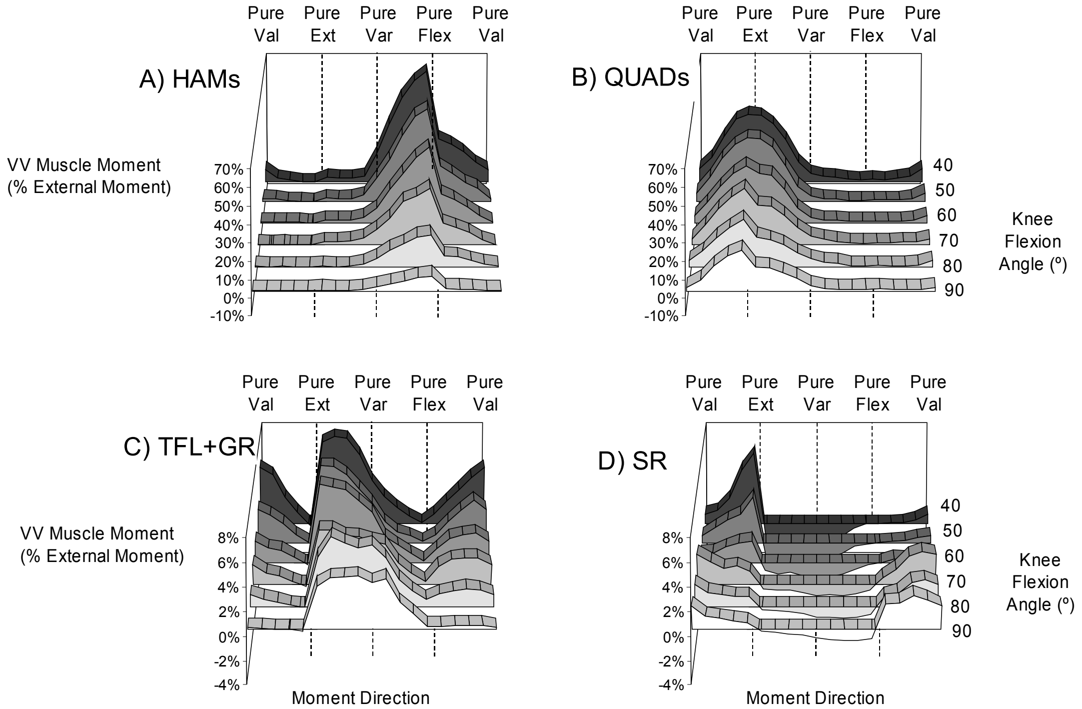

Figure 3 - Muscle contributions (% external moment) to the external varus and valgus moment for A) hamstrings, B) quadriceps, C) tensor fascia latae and gracilis, and D) sartorius. These are plotted verses knee joint angle and moment direction. NOTE, the change in y axis scale on C) and D).

their VV moment arms and summing the subsequent individual muscle VV moments.

The *residual load* constitutes the potential for soft tissue loading. The residual load is the difference between the internally generated muscle moments (determined from the model) and the external applied moments (determined using inverse dynamics). The *residual FE load* is by assumption equal to zero. However, the *residual VV load* is determined by subtracting the externally applied VV moment from the internal VV moment generated by the muscles. If the VV moment generated by muscles is *greater than* that applied externally, then there is *no* residual load and *no* potential for soft tissue loading. If, however, the VV moment generated by muscles is *less than* that applied externally, then there is a residual VV load and thus potential for soft tissue loading. The *residual load ratio* is defined as the residual VV load expressed as a percentage of the magnitude of the combined FE and VV external load.

Static muscular support of varus-valgus knee moments. The results showed that muscles were primarily used to support flexion and extension loads at the knee, but in so doing, were able to support some part of the varus or valgus loads (Fig. 3). However, soft tissue loading was still required. Soft tissues supported up to an average maximum of 83% of the external load in pure varus and valgus. Soft tissue loading in pure varus and valgus was less than 100% of the external load as the muscles, on average, were able to support 17% of the external load.

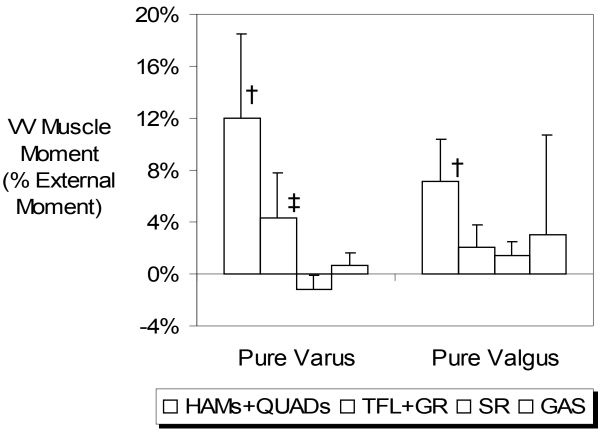

Figure 4 - The VV muscle group moments (% external moment) during pure varus and pure valgus loads. The values are the subject averages and standard deviations averaged across knee joint angle. † HAMs+QUADs VV moments significantly larger (p<0.001) than each of the SR, TFL+GR, and GAS: ‡ TFL+GR VV moments significantly larger (p<0.001) than each of the SR and GAS.

The hamstrings and quadriceps were specifically activated to support the flexion or extension moments respectively (see figure 3A and 3B). However, it was found that the 1) co-contraction of the hamstrings (HAMS) and quadriceps (QUADS), and 2) activation of the gracilis (GR) and tensor fascia latae (TFL) were more tuned to the magnitude of the varus and valgus moments (Lloyd and Buchanan, 2000; Buchanan and Lloyd, 1997). The hamstrings and quadriceps supported most of the varus and valgus moments (Figure A & B). In pure varus and pure valgus hamstrings and quadriceps co-contraction supported 8% to 12% of the external moment (Lloyd and Buchanan, 2000). The sartorius (SR) had negative contributions during varus loads, but had positive contribution during valgus loads. The gastrocnemus (GAS) contribution was always low, which was probably because the feet were flail during the tasks. There were definite activation strategies to support varus and valgus moments, albeit small, which suggest dual goals of the neuromotor system to support varus and valgus moments.

APPLICATION OF THE DYNAMIC MODEL: Muscular support of varus and valgus loads at the knee during running and cutting. The purposes of this study were two fold: 1) to determine if the dynamic version of the EMG driven model of the human knee (as presented above) could be used to accurately and reliably estimate knee moments across a varied range of dynamic contractile conditions, and 2) to determine muscle contributions to dynamic varus and valgus loading.

Six subjects were tested (mean age: 20.5 ± 2.9 years; mean mass: 74.6 ± 8.6 kg) and 4 of these 6 subjects were tested one week later to test the reliability of the model across weeks. Subjects performed a series of tasks on a Biodex isokinetic dynamometer (Shirley, NY) and a series of running and sidestepping manoeuvres in a 3 dimensional gait laboratory.

The Biodex tasks included: maximum isometric efforts, passive FE; eccentric hamstring and quadriceps contraction, low effort concentric FE; and maximal effort concentric FE. During these trials, knee FE torque, knee flexion angle, and EMG data from 10 knee muscles were collected at 2000 Hz.

The subjects performed a series of running and cutting manoeuvres at ~3 m/sec, the latter being a dynamic challenge to knee joint stability. The cutting tasks were sidesteps to 60° and 30° from the direction of travel, and a crossover cut to 30° from the direction of travel. Lower limb joint kinematic data were collected with a 6-camera 50 Hz VICON Motion Analysis system (Oxford Metrics Inc.) using a VICON Clinical Manager (VCM) marker set (Kadaba *et al.*, 1990). Force data were collected simultaneously at 2000 Hz using an AMTI force plate, and input into an inverse dynamic model to calculate knee FE moments across the stance phase for each manoeuvre (Kadaba *et al.,* 1990). EMG data from the same 10 knee muscles were also collected at 2000 Hz.

The model was calibrated for each subject using 5 trials. The calibrated model was then used to predict net FE muscle moments across all tasks and compared to the inverse dynamics FE joint moments for validation.

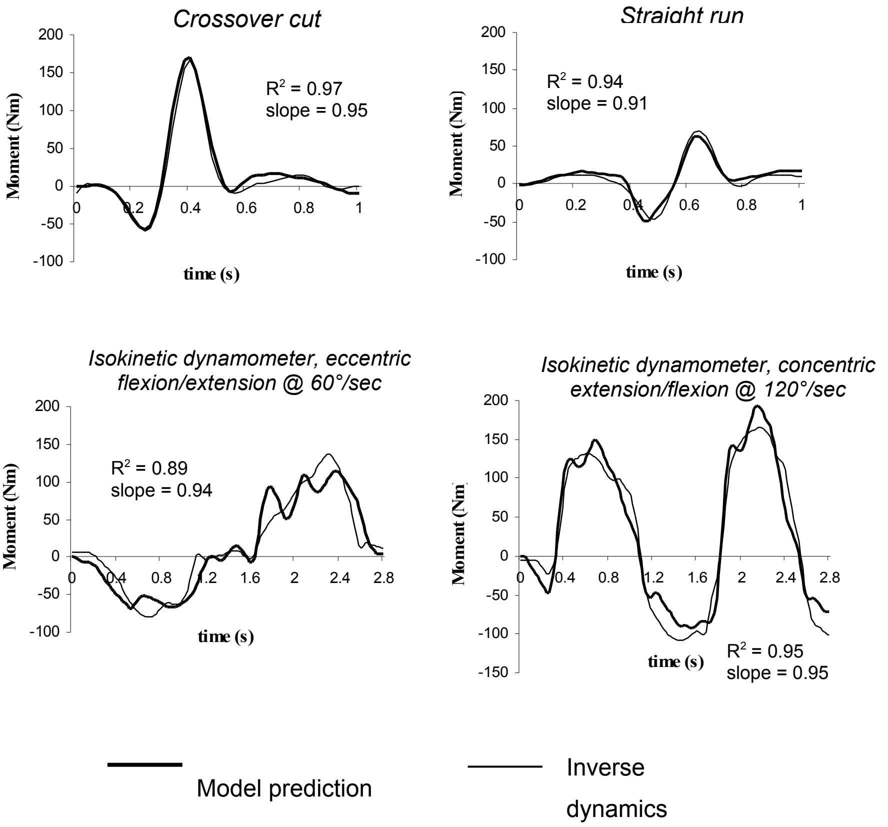

Figure 5 - Comparison of model predictions of knee FE muscle moments with FE knee joint moments measured externally for subject 1. Extension moments are positive on all graphs.

Model performance. Following calibration, the model was able to predict FE knee moments with a mean (S.D.) coefficient of determination (R^2) of 0.91 ± 0.04 across 204 running, sidestepping and dynamometer trials. Mean residual error for these predictions was ~8 Nm and when normalised to body weight was less than 0.03 Nm/kg. The dynamic knee model was capable of predicting FE moments across a wide range of tasks from running, to crossover cutting and eccentric dynamometer tasks (Figure 5).

The model was retested for four subjects who returned for a repeat testing session one week later. Using the muscle model parameters from the first weeks' calibration, but recalibrating the EMG-to-activation coefficients, there was no decline in the model prediction accuracy of the FE joint moments (First week: R^2= 0.91 ± 0.018; Second Week: R^2= 0.91 ± 0.031).

Dynamic muscular support of varus and valgus knee moments. The residual loads were calculated the same as for the static case. This type of analysis does not consider residual loads in the internal-external rotation and anterior-posterior draw, thus under estimate the possibility of soft tissue loading.

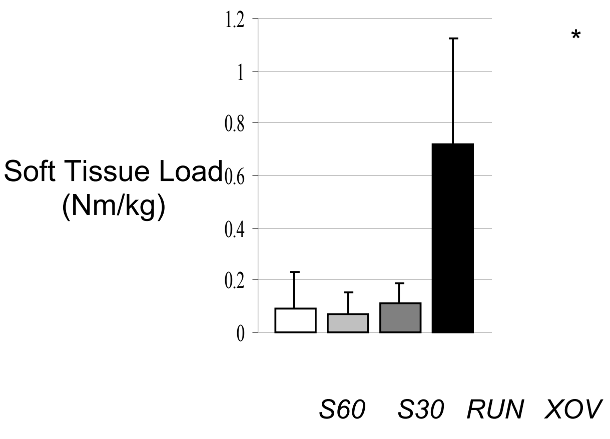

Figure 6 - Peak residual load during running and sidestepping tasks. The crossover cut (XOV) task potentially placed significantly more load on soft tissues than the other tasks (indicated by the *, p < 0.01).

The residual VV loads measured across the stance phase during the running and cutting tasks demonstrate that muscles were capable of resisting large VV external loads applied to the knee joint during the dynamic functional tasks (see Fig. 6). The residual VV loading was greatest for the cross over cut.

Comparing static and dynamic muscular stabilisation of the knee joint. During the sidestepping tasks (S30 and S60) the residual load ratios were 1%, the muscles being capable of resisting 99 % of the combined FE and VV external load applied to the joint. During the RUN and crossover cut tasks the residual load ratio was also only 1 % and 5 % respectively. In comparison, in the static joint stabilisation tasks the residual load ratio was 50% when knee had similar relative external and flexion angle to that seen in the running and cutting tasks. These results suggest that muscles relative contribution to the dynamic VV knee stabilisation is far greater than that observed during static joint stabilisation.

CONCLUSIONS: The static and dynamic EMG driven models we have developed can estimate muscle forces at the knee validated against the model's ability to predict the knee FE moments during a range of static and dynamic tasks. The validity and reliability of the model makes it a useful tool for investigating the potential for soft tissue loading during tasks that challenge knee stability. This can then be used in identifying injury mechanisms and risk of injury when performing common sporting manoeuvres. The models can also used to assess changes that occur during neuromuscular training studies since the model implicitly incorporates activation patterns and the changes that may occur due to exercise.

Model improvements. We are currently incorporating internal-external rotation and anterior-posterior draw as additional degrees of freedom in the anatomical knee model so muscle contribution to these movements can be predicted. We are also developing anatomical models that incorporate ligaments, cartilage, and menisci. Muscles will also be modelled as multiple lines segment, especially for the vastus lateralis and vastus medialis, which originate over large areas on the femur.

Also currently being tested are different forms of the EMG to activation model and the Hill-type muscle model. Specifically, we have been testing different ways to model the EMG to activation non-linearities, the time delay between activation and force, velocity history, and cross-talk. We have also been examining the use of first order differential equations versus discrete second order differential equations to characterise the muscle activation dynamics. Discrete versions of the muscle activation dynamics mean that these can be implemented in digital signal processing hardware for more rapid calculation of activation. Modifications to the Hill-type muscle model are being evaluated such as making the maximum contraction velocity dependent on activation and muscle fibre length (Hatze, 1977), and linking muscle fibre force output to work history (Herzog, 1998).

Model use. If this type of modelling is to be more routinely and widely adopted in biomechanical and epidemiological studies, then it must be simple and rapid in its use. At the moment the calibration is time consuming, ranging from 2hrs to 72hrs (on a Silicon Graphics R10000 O2) depending on the number of calibration parameters included in the model. However once calibrated, the model could be implemented to work in real time.

EMG driven models have been used sparingly but very effectively to date. With future refinement of such models and their wide spread adoption, investigations of in-vivo joint articular surface load, examination of the energy of flows across joint during different movements, or the in-depth study of the stretch-shorten cycle, will be common place. *Only then can we start answering the questions on why and how does tissue-loading lead to injury or disease during physical activity.*

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