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Could Passive Knee Laxity be Related to Active Gait Mechanics? An Exploratory Computational Biomechanical Study Using Probabilistic Methods

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Abstract

Higher expectations and increasing demand for total knee replacement (TKR) necessitates better understanding of the many factors influencing clinical outcomes. Recently, probabilistic methods have been used to model the influence of variability on TKR mechanics. This study demonstrates conceptually how probabilistic might provide a framework to explore relationships between different activities, e.g. passive laxity range-of-motion tests (as can be performed intra-operatively) and post-operative active gait mechanics. Two implants were compared using simulated ISO-gait and passive laxity tests, with input factors including mal-positioning and soft-tissue constraint varied using a Monte-Carlo model. The results illustrate that correlations between different activities can be quantified; these correlations are design-dependent and of varying strength; nonetheless the existence of correlations in this demonstration study suggests further research is needed (with detailed clinically-representative models) to explore the relationship between passive and active mechanics for specific *in-vivo* conditions. Probabilistics is a key enabling methodology for achieving this goal.

Keywords

Knee Biomechanics, Computational Wear Modelling, Probabilistic Methods

Introduction

Demand for TKR continues to grow, whilst simultaneously, diversifying patient demographics [1] and high patient expectations [2] present challenges to established TKR technologies and procedures. Therefore, in order to improve pain relief, function & longevity, efforts are ongoing to lower failure rates by improving both design and clinical practice.

A large number of studies have been published which consider individual parameters to determine their influence of TKR outcomes, e.g. specific mal-orientations or eccentric loading [3-5]. However, these studies often treat factors in isolation. Probabilistic studies, which attempt to model the combined influence of a wide range of factors, have been adapted from structural engineering applications to be applied to orthopaedic devices [6, 7]. More recently, probabilistic models have been introduced specifically to TKR mechanics [8, 9]. The advantage of these models is the ability to explore more completely the 'possibility space' created by the inherent variability in various input factors (e.g. component positioning or material properties). By integrating all the significant input factors into a single model, it is possible to explore how different input factors and output responses interact together, in a statistically quantitative framework.

The number of non-identical trials required for a probabilistic study means that computer-assisted simulation is generally the most appropriate modelling method. Full deformable finite-element simulations are still too computationally expensive, taking several hours to compute [10], so fast rigid-body models are preferred [11]. Developments in computer technology now permit high-volume studies with hundreds or even thousands of trials to be performed within the a reasonable timescale [12, 13].

However, such studies can generate copious volumes of numerical data, and a key challenge to overcome before probabilistic models are widely adapted is how to condense this data into a concise, readily-interpretable format for the benefit of designers or clinicians. An important indication of this practical relevance is whether probabilistic studies can provide a framework to derive simplified, generalised rules and relationships which can be more readily interpreted and applied. To begin addressing this issue, this study presents conceptually how probabilistic methods might be used to identify and quantify such relationships between two output performance tests: the 'passive' laxity motions of the knee, and the kinematics & peak contact pressures experienced in an 'active' gait cycle.

Simple passive laxity tests can readily be performed intra-operatively, but the question of whether these tests can yield information about the likely post-operative 'active' performance of the knee has yet to be rigorously addressed. Currently, this is a subjective judgement based on the expertise of the clinical professional. A comparison using simulation methods may allow more quantitative statements to be made about the predictive power (and hence practical value) of these passive laxity tests. This paper will demonstrate how such a study might be structured, using simplified computational simulations of simulated gait and laxity draw tests.

Methods

This study is based upon an adaptation of the published probabilistic investigation by Laz et al [9], and incorporates several variables included in that study (mal-positioning, mal-rotation, friction and medial-lateral loading input factors). However, various developments are introduced in order to explore the passive/active performance correlations.

The baseline computational model used for this study is a rigid-body simulation running in MSC.ADAMS (MSC.Software Corporation), as reported previously [14, 15]. For this study, a cruciate-retaining fixed-bearing TKR design was modelled. To differentiate design-specific correlations, two alternative tibial inserts were simulated for comparison: a semi-constrained (S/C) design and an unconstrained (U/C) design (with reduced conformity in the sagittal plane).

The gait cycle mechanical configuration is a derivative of the force & displacement-driven Instron/Stanmore knee wear simulator [16]. However, in order to simulate laxity effects, the transverse-plane spring restraint system was replaced with an elemental ligament restraint system, composed of three non-wrapping nonlinear spring elements representing the posterior cruciate ligament and the two collateral ligaments. The force (F) – strain (ε) relationship was adopted from previous analytic studies [17] as shown in Eq. 1:

$$F = \begin{cases} 0 & \varepsilon \leq 0 \\ \left(\dfrac{k}{4\varepsilon_1}\right)\varepsilon^2 & 0 < \varepsilon \leq 2\varepsilon_1 \\ k(\varepsilon - \varepsilon_1) & 2\varepsilon_1 < \varepsilon \end{cases}$$

There are three controllable parameters within this ligament model: linear-region stiffness (k), toe-in (ε_1), which determines the strain level at which the linear region begins, and pre-strain (ε_p), which determines pre-loading in the ‘neutral’ (knee extended) position. Variability levels for these factors were derived from the literature [18]. Although multi-bundle models have been shown to better represent the ligaments [19], single unified models were chosen for this study to limit the number of input variables under investigation, with the three variables (ε_1, ε_p & k) for each of the three ligament elements giving a total of nine additional factors.

Combined with the factors adopted from published studies (using the higher levels of variability reported), the complete set of input variables are listed in table 1; all variables are assumed to be independent, following a Gaussian distribution bounded at ±3 standard deviations.

As in previous studies, output kinematics and peak contact pressures were analysed through a standard gait cycle, based on ISO-derived force/displacement input waveforms [10] – see figure 1. For the gait cycle, the selected output measures were A-P translation & I-E rotation, and peak contact pressure, sampled throughout the gait cycle. Kinematics are reported in terms of ‘offset’ values; i.e. normalised relative to the equilibrium position at 0% gait cycle.

Additionally, three pairs of passive laxity draw tests were simulated with typical clinical passive loading levels [20]: anterior-posterior (A-P) draw (±100 N), internal-external (I-E) torsion (±5 Nm), and varus-valgus (V-V) torsion (±10 Nm). Laxity tests were simulated both in full extension and at 20° flexion (reflecting the clinical practice of testing at high-laxity flexion angles), with compressive axial loading limited to 300N for ‘passive’ restraint [21]. In each case the output measure was the offset displacement motion in the direction of the applied draw loads.

In order to provide a matched set of trials (so as to directly compare correlations), search-based fast-probability integration methods were not used; instead, a 1000-trial matched Monte-Carlo analysis was performed, with the same matrix of input factor settings used for both the gait cycle and laxity draw simulations. Because this simplified model did not include capsule or musculature contributions to joint restraint, the range and levels of variability studied meant a handful of statistically outlying trials resulted in subluxation under high laxity draw tests. These trials were excluded from the subsequent correlation analysis.

Results from the simulations were used to determine 1%-99% performance envelopes for gait cycle kinetics & kinematics, and to determine the statistical distributions for laxity test displacements for both designs.

To identify correlations between active gait and passive laxity, scalar statistical metrics for the time-varying gait waveforms were required for each trial. The waveform minimum, maximum, mean, range, and standard deviation were chosen for this purpose. Each of these five values was calculated for the three gait cycle output measures, and the results from all trials correlated against the three pairs of laxity test displacements, giving a 3×15 correlation matrix. This matrix was generated for both the S/C and U/C designs to allow comparison between designs.

Results

Active gait simulation characteristics

Probabilistic performance envelopes for the simulated ISO-wear gait cycle were calculated for comparison against previous studies. The gait kinematics and peak contact pressure are shown in figure 2. Both S/C and U/C designs are included on the same axes for comparison; the kinematics are reported as ‘offset’ values to more clearly illustrate design-specific differences between S/C and U/C.

For S/C, it is apparent that the offset A-P and I-E motion of the knee is much more closely constrained to the 'neutral' (unperturbed) motion during stance phase, whereas the U/C design permits more variability of motion in stance phase. For both designs, the variability envelope expands to its widest during swing phase (due to lower articular surface conformity and lower compressive forces). The differences between the S/C and U/C designs are also apparent in general envelope trends, with larger envelopes for the U/C design (reflecting the larger mean-value kinematics and contact pressures).

Passive laxity draw-test correlations

The distributions of laxity draw range for A-P, I-E and V-V are shown in figure 3. The range is the full difference in displacement between the two opposite draws; e.g. between the displacement for posterior draw of -100N, and the displacement for anterior draw of +100N. Again, both S/C and U/C results are plotted together, to illustrate the design differences. It is clear that the most significant difference between designs is for the A-P draw tests, where the lower sagittal conformity of the U/C design allows higher draw ranges. The distribution of I-E laxity for the two designs lie within a similar range, whilst for V-V laxity, the distributions are very similar in shape, with higher laxity for the S/C design.

Passive-active correlations

The correlations are reported in terms of Pearson-squared (R^2) values in table 2, with the strongest correlations highlighted. Note that these values indicate the strength of the correlation only (and not whether a correlation is positive or negative). The correlation coefficients are mostly low; this is to be expected of a complex mechanical system with multiple influential factors. Nonetheless, some of the correlations are sufficient to provide some degree of predictive power (R^2 up to ~ 0.5). Notable trends are apparent for both the S/C and U/C designs; especially for the I-E and V-V torsional tests. The largest difference between the two designs was for A-P draw tests; this would be expected, since the design difference between the two is in the sagittal plane, and would most directly affect A-P laxity. The more conforming S/C design showed limited predictive power between active & passive mechanics (with laxity motion restricted), whereas the U/C design had higher correlation coefficients.

For active gait parameters, the strongest correlations occurred for contact pressures, with moderate correlations for I-E rotation and very little correlation for A-P translation. The laxity tests had greater predictive power for variations in minimum (i.e. swing phase) contact pressures, suggesting that it is the influence of the modelled ligaments which is causing this correlation. I-E rotation was not well-correlated for the S/C design; only small I-E rotations occur for this design during gait, so any correlations are likely to be less evident. For illustrative purposes, figure 4 provides representative example correlation scatter plots for the weakest and strongest correlations observed.

Discussion

The performance envelopes predicted for the normal gait cycle are comparable to published studies under similar conditions [9]. A larger degree of output variability is evident in the present study, due to the additional input variability factors related to the 'ligament' spring-elements. This is most apparent in the peak contact pressure envelopes, where a variability range of up to 6MPa is seen for both designs. The effect of compressive forces due to the ligament elements will increase contact pressure ranges; this is further compounded by mal-positioning leading to exaggerated gait kinematics with lower tibiofemoral conformity at contact, and hence higher contact pressures.

When the envelopes for S/C and U/C designs are compared, these results do suggest that insert design can play a role in controlling the influence of variability on gait mechanics. However, this may be specific to the simplified mechanical configuration being demonstrated in this conceptual study, and further studies using more extensive models would be required to confirm this observation.

The laxity tests reveal a high degree of torsional laxity in these simulations at both flexion angles (0° and 20°); this must be interpreted in light of the reduced transverse plane restraint provided by the ligament elements used in the model (comparable cadaver experiments using this tibiofemoral test configuration have also yielded high levels of rotation [22, 23]). Trends were similar at both flexion angles simulated, with (as would be expected) generally greater magnitudes of laxity range measured for the 20° position. As anticipated, the most apparent differences between designs were for the A-P laxity tests; it is notable that when the full distribution of variability is considered, higher V-V laxity is evident for the more constrained S/C design than the U/C (this is associated with greater I-E rotation of the tibial component under V-V torques for the S/C compared to U/C designs, suggesting that this rotation may facilitate the higher V-V laxity without requiring condylar lift-off).

Although the range of passive laxity motion is very low for V-V rotation, the correlations are generally strongest; this is most likely because much of the correlation is due to variability in the simplified ligament elements. Whereas A-P and I-E tests are in the transverse plane (normal to the principal ligament orientation and requiring more displacement to recruit the ligaments), V-V tests can directly distract the joint, resulting in increased loading of the ligaments and so giving more ready indication of variability in the soft-tissue constraint. Note that this is contrary to the configuration of many knee wear simulators, where spring restraint is applied only in the transverse plane; again, illustrating the influence that the choice of mechanical model may have upon the study outcomes.

This conceptual study has explored the relationship between the influence of variability on passive laxity and gait kinematics & kinetics, for two specific TKR design variations, using a simplified mechanical model of the tibiofemoral joint with rudimentary ‘soft tissue’ representation. Correlations were demonstrated for certain parameters – in some cases, with predictive powers up to R^2=0.5. This may allow design-specific predictions about gait mechanics to be made based on laxity tests; for example, high V-V laxity means it is more probable that a knee with the U/C insert will experience greater I-E rotation in gait; the same trend is less probable for the S/C insert. This becomes clinically relevant when these mechanical observations are related to modes of failure; for example, studies have associated more pronounced I-E rotational kinematics with higher component wear [24, 25]. It could therefore be hypothesised that the U/C implanted knee exhibiting higher passive V-V laxity might experience greater wear damage from prolonged active gait; however, the same would not be expected for the S/C implanted knee, where the correlations are weaker.

This study is intended only to illustrate the use of statistical correlations to link the characteristic mechanics of different active and passive daily activities, and there are some important limitations to the models which should not be overlooked. The simplified ligament model and rigid-body contact formulations will result in reduced accuracy, and the set of input variables studied is limited, with generalised assumptions made about distribution form and correlations which would need to be better modelled. The opportunity exists to develop this methodology with more complete and accurate anatomical models to explore (for specific designs) whether passive laxity can be a predictor for active gait mechanics. The results of this exploratory study suggest this may be the case, for a limited sub-set of gait characteristics, and subject to design-dependency. It remains to be investigated whether other activities (e.g. stair usage or deep flexion manoeuvres) would exhibit similar correlations to passive laxity motion, for a range of different flexion angles.

It would require further investigation with a wider range of variability factors and implant designs to determine how much of this correlation is universal (i.e. related directly to the variable input factors themselves, such as ligament effects), and how much is controlled or constrained by the implant design (such as S/C A-P motion in this study). However, it seems apparent that certain design features (e.g. lower constraint) can improve predictive power, and that some laxity tests (e.g. V-V) correlate better than others to gait characteristics. Measurement errors are known to be associated with measurements of passive laxity [26]; this would erode the strength of these correlations, so a more exhaustive study would also need to account for uncertainty in the laxity ranges.

It is particularly important for probabilistic studies to clearly and succinctly consolidate the large volume of data generated, to ensure the results are concise, relevant and accessible to orthopaedic designers or clinicians. Although the amount of data generated by probabilistic methods such as this present study can be extensive, it is possible to condense this information down to more useful and meaningful metrics. Representing results with variability envelopes and sensitivity factors is one established approach; summarising correlations between different activities may be another practical method, allowing another perspective on the mechanical interactions of the system, and providing ‘rule of thumb’ estimates for co-relational behaviour across different activities.

Probabilistic methods are still relatively underexploited in the field of biomechanics; in order to become more widely adopted they must demonstrate their relevance and accessibility to the research & design community.

This study illustrates conceptually another potential application of probabilistics, demonstrating the design-dependent correlations between passive laxity and active gait mechanics, and suggesting that for some gait characteristics these correlations potentially offer useful predictive power as a decision-support tool.

Acknowledgments

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Tables

<table>
<tr><th>Factor</th><th>Mean (μ)</th><th>St.Dev (σ)</th></tr>
<tr><td>Fem. axis I-S</td><td>25.4 mm</td><td rowspan="4">0.5 mm</td></tr>
<tr><td>Fem. axis A-P</td><td>0 mm</td></tr>
<tr><td>Tib. axis M-L</td><td>0 mm</td></tr>
<tr><td>Tib. axis A-P</td><td>7.62 mm</td></tr>
<tr><td>Initial flexion°</td><td rowspan="4">0</td><td rowspan="4">1°</td></tr>
<tr><td>Initial I-E°</td></tr>
<tr><td>Initial tilt°</td></tr>
<tr><td>Initial V-V°</td></tr>
<tr><td>Friction coefficient</td><td>0.04</td><td>0.01</td></tr>
<tr><td>M-L load split</td><td>60M-40L</td><td>2.5%</td></tr>
</table>

<table>
<tr><th>Factor</th><th>Mean (μ)</th><th>St.Dev (σ)</th></tr>
<tr><td>LCL k</td><td>70 N/mm</td><td rowspan="3">20%</td></tr>
<tr><td>MCL k</td><td>100 N/mm</td></tr>
<tr><td>PCL k</td><td>130 N/mm</td></tr>
<tr><td>LCL ε_p</td><td>+5%</td><td rowspan="3">1%</td></tr>
<tr><td>MCL ε_p</td><td>+0%</td></tr>
<tr><td>PCL ε_p</td><td>+2%</td></tr>
<tr><td>LCL ε_1</td><td rowspan="3">+3%</td><td rowspan="3">1%</td></tr>
<tr><td>MCL ε_1</td></tr>
<tr><td>PCL ε_1</td></tr>
</table>

Table 1. Input variables under study, with mean and standard deviation values

		S/C			U/C		
		A-P	**I-E**	**V-V**	**A-P**	**I-E**	**V-V**
GAIT AP	**MIN**	0.05	0.03	0.01	0.12	0.04	0.00
	MAX	0.01	0.00	0.03	0.02	0.00	0.05
	RANGE	0.05	0.08	0.12	0.10	0.06	0.07
	MEAN	0.02	0.00	0.01	0.07	0.01	0.01
	ST.DEV	0.07	0.04	0.02	0.06	0.04	0.08
GAIT IE	**MIN**	0.01	**0.12**	0.05	**0.15**	0.08	0.00
	MAX	0.01	0.00	0.09	0.02	0.02	**0.21**
	RANGE	0.01	**0.15**	0.10	**0.20**	**0.36**	**0.34**
	MEAN	0.05	0.09	0.04	**0.16**	0.07	0.00
	ST.DEV	0.03	**0.12**	0.09	**0.25**	**0.38**	**0.33**
GAIT CP	**MIN**	0.08	**0.20**	**0.31**	**0.23**	**0.31**	**0.42**
	MAX	0.00	0.01	0.00	0.09	0.02	0.02
	RANGE	0.03	0.10	0.10	**0.24**	**0.13**	0.03
	MEAN	0.04	**0.36**	**0.36**	0.06	**0.27**	**0.33**
	ST.DEV	0.06	**0.43**	**0.40**	**0.41**	**0.52**	**0.33**

Table 2. Correlation matrix: active gait parameters (rows, headings left) versus passive laxity draw ranges (columns, headings top) for S/C & U/C designs

Figure legends

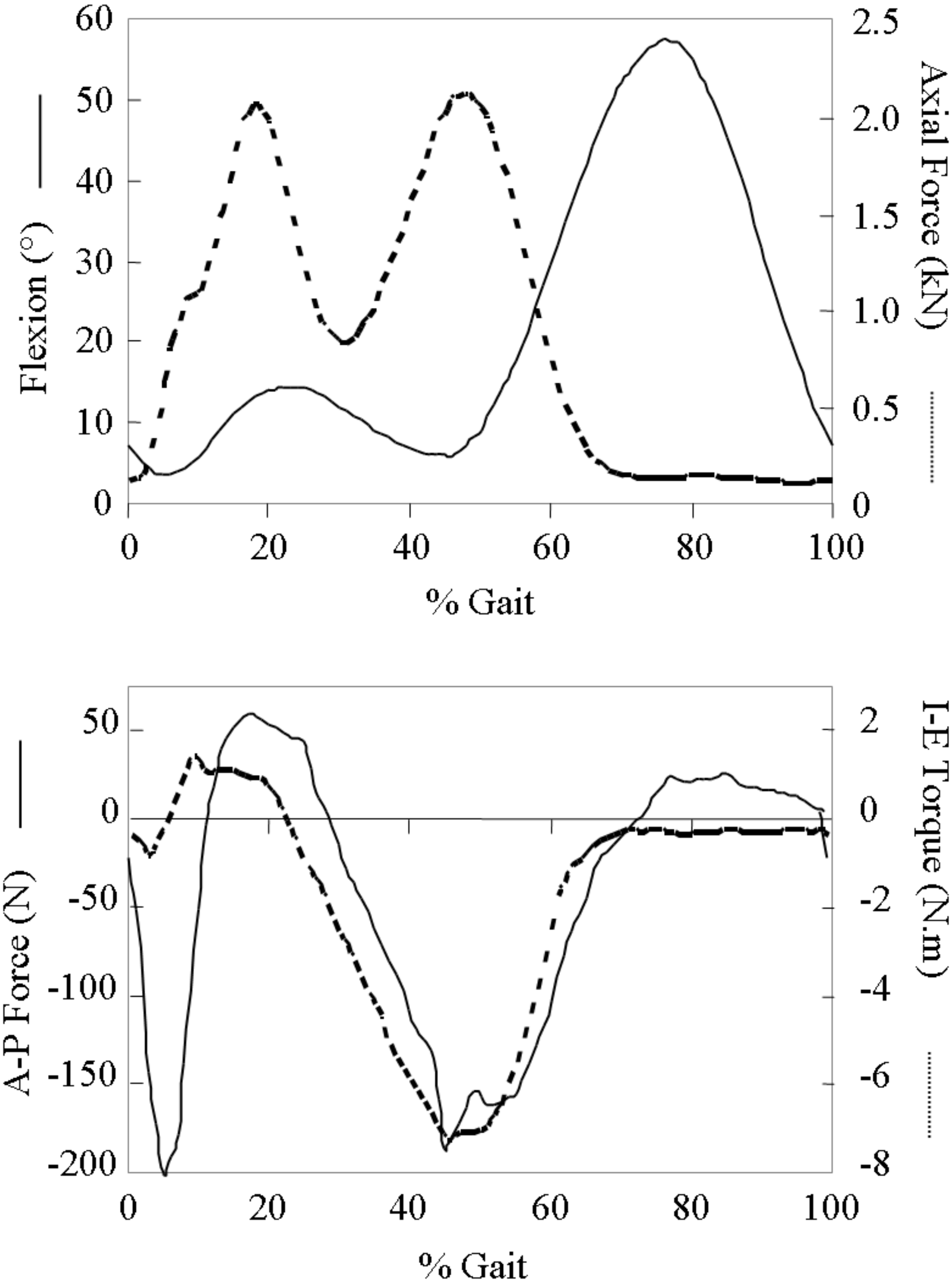

Figure 1. Input waveforms for force-driven gait simulation

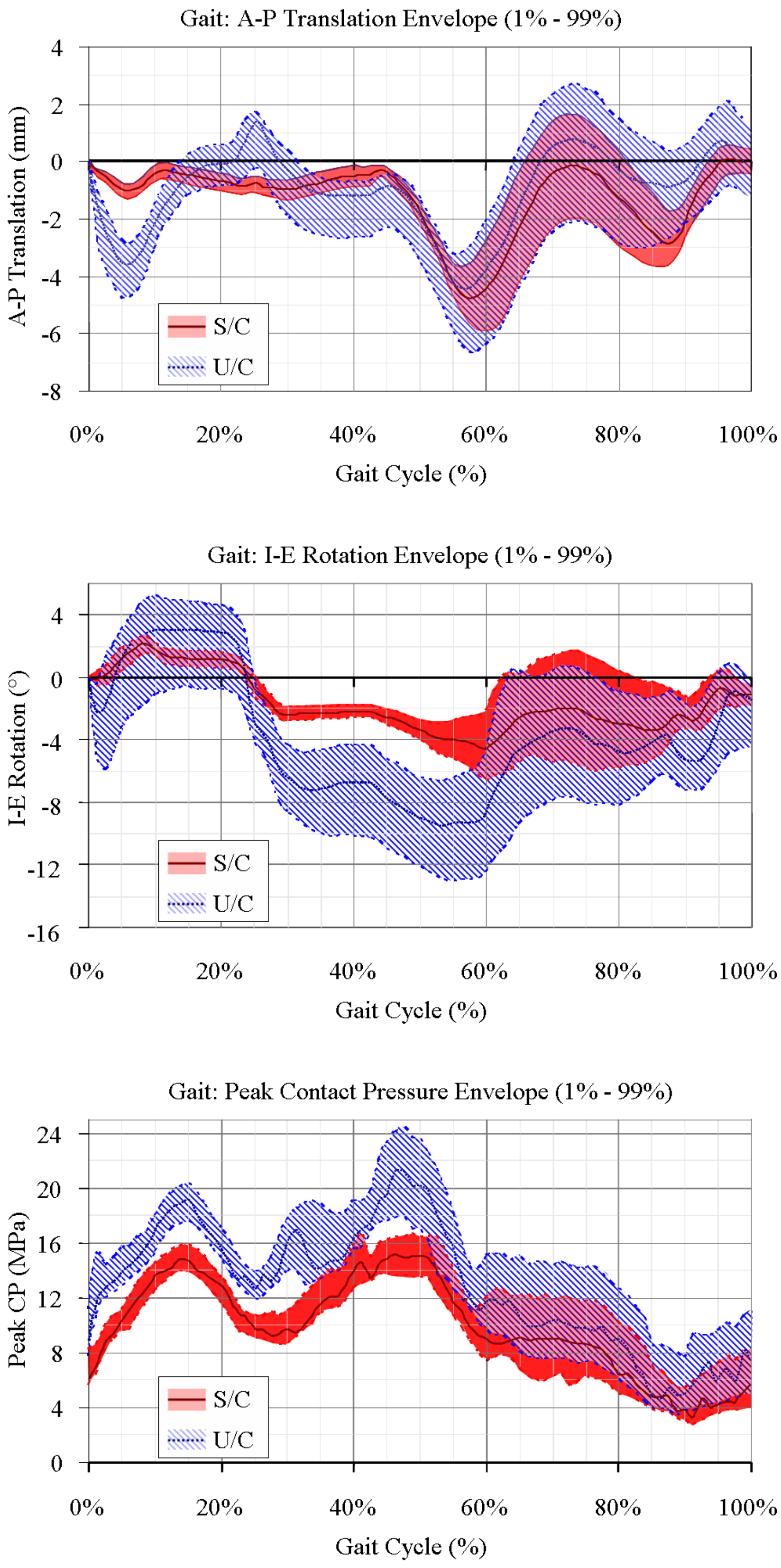

Figure 2. Kinematics and peak contact pressures for gait simulation, with variability envelopes. The deterministic 'neutral' case shown is a solid bold trace, whilst the semitransparent envelope represents the range of values with a probability of occurrence between 1% and 99%

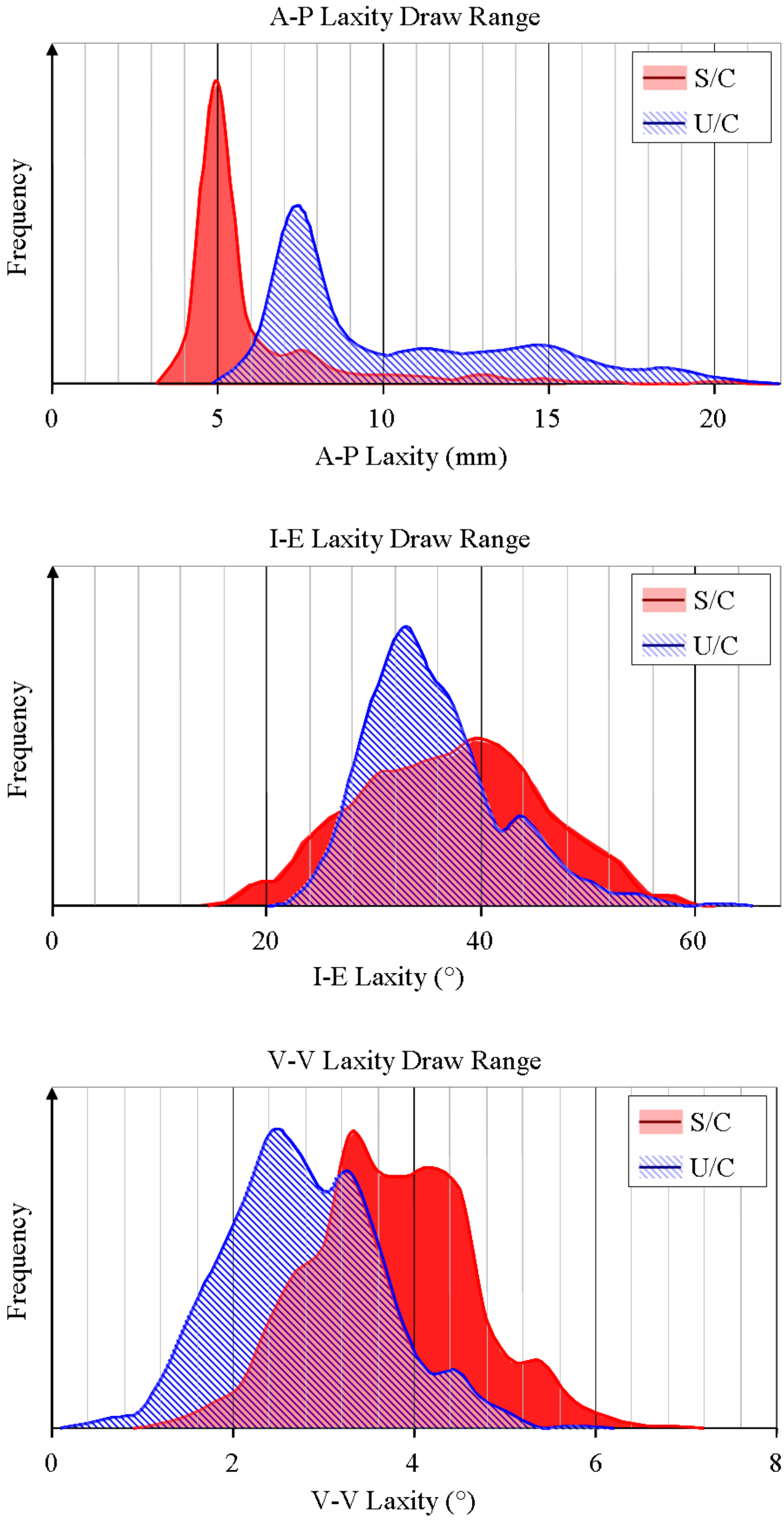

Figure 3. Distribution in laxity draw ranges due to input variability

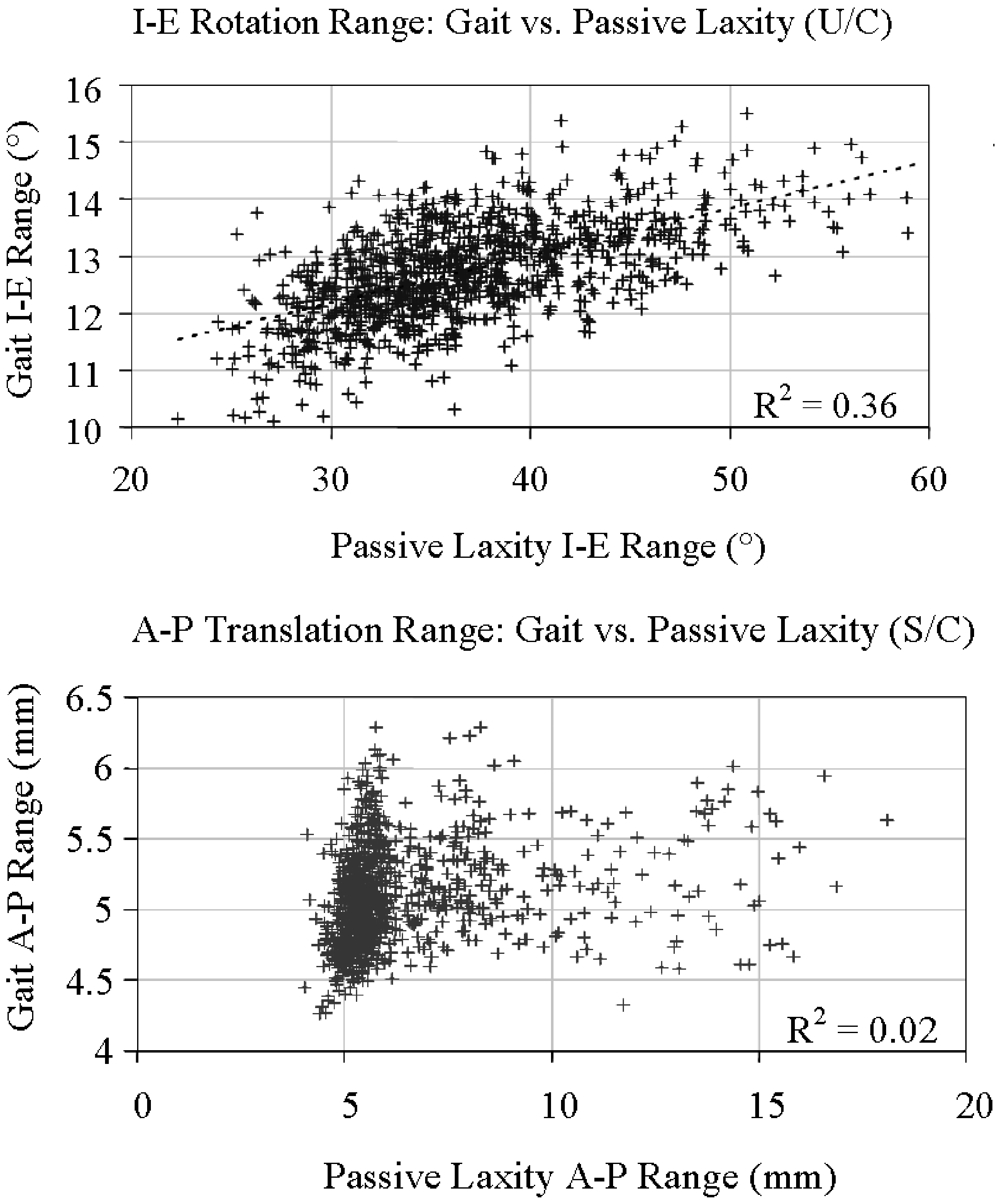

Figure 4. Examples of correlations: strong (I-E range for U/C) & weak (A-P range for S/C)