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The role of the heel pad and shank soft tissue during impacts: a further resolution of a paradox

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ABSTRACT

The aim of this study was to test the hypothesis that by accounting for soft tissue motion of the lower leg during the impacts associated with *in vivo* testing, that the differences between *in vivo* and *in vitro* estimates of heel pad properties can be explained. To examine this a two-dimensional model of the shank and heel pad was developed using DADS. The model contained a heel pad element and a rigid skeleton to which was connected soft tissue which could move relative to the bone. Simulations permitted estimation of heel pad properties directly from heel pad deformations, and from the kinematics of an impacting pendulum. These two approaches paralleled those used *in vitro* and *in vivo* respectively. Measurements from the pendulum indicated that heel pad properties changed from those found *in vitro* to those found *in vivo* as relative motion of the bone and soft tissue was allowed. This would indicate that pendulum measures of the *in vivo* heel pad properties are also measuring the properties of the whole lower leg. The ability of the wobbling mass of the shank to dissipate energy during an impact was found to be significant. These results demonstrate the important role of both the heel pad and soft tissue of the shank to the dissipation of mechanical energy during impacts. These results provide a further clarification of the paradox between the measurements of heel pad properties made *in vivo* and *in vitro*.

KEYWORDS: heel pad, wobbling mass, simulation

INTRODUCTION

During walking and running there is a significant impact force associated with each footfall (e.g. Light et al., 1980; Dickinson et al., 1985). The human heel pad is assumed to be responsible for the dissipation of some of the mechanical energy associated with initial foot contact. Measurements of heel pad properties made *in vivo* and *in vitro* show significant differences, which makes the determination of the role of the heel pad in the dissipation of energy difficult to quantify. In addition there is evidence that the soft tissue of the human body moving independently of the skeleton may also have a role in energy dissipation (e.g. Gruber et al., 1998). This study aims to examine the relative contributions of soft tissue motion and the heel pad to energy dissipation during impacts.

Examination of the mechanical properties of the human heel pad has taken two routes. *In vivo* testing by way of impact experiments (e.g. Cavanagh et al., 1984; Kinoshita et al., 1993; Valiant, 1984; Nigg et al., 1984), and *in vitro* testing, primarily Instron testing (e.g. Noe et al., 1993; Bennet and Ker, 1990; Aerts et al., 1995). *In vivo* testing has attempted to reproduce controlled impacts similar to those found in jogging. These *in vivo* experiments have initial impact velocities of around 1 m/s and involve a collision between the heel pad and an instrumented rigid material. The lower leg is considered to be fixed and immovable during these tests. Force-deformation curves are then measured to determine stiffness values and energy losses due to damping of the heel pad. These methods invariably ignore the lower leg, foot and supporting framework as variables in the experiment. Valiant (1984) assumed that the leg was rigidly fixed and the pendulum impacts acted through the axis of rotation of the ankle joint, allowing recording of the kinematics and kinetics of the heel pad only. Even if these assumptions hold, the properties of the heel pad alone are probably not measured because of relative motion between the soft tissue and the underlying bone that is seen to some extent in all impacts (e.g. Cappozzo et al., 1996; Fuller et al., 1997). The fact that *in vitro* and *in vivo* testing give different results would tend to support this idea. *In vivo* measurements typically give heel pad stiffness values of approximately 150 kN/m at one bodyweight loading, exhibit energy losses up to 95%, have peak force values of 800 N, and are frequency dependent (Cavanagh et al., 1984; Valiant, 1984). These *in vivo* results have been used to model the foot-ground interface in biomechanical models (e.g. Gilchrist and Winter, 1996; Wright et al., 1998; Güler et al., 1998).

In vitro experiments utilizing Instron testing on the heel pad and part of the calcaneous have obtained different results compared with those from *in vivo* experiments. The stiffness at one bodyweight loading is between 1160 kN/m and 1445 kN/m (Bennet and Ker, 1990; Aerts et al., 1995), energy loss is only about 30%, peak force values of 1800 N, and the results are not frequency dependent.

Aerts et al. (1995) referred to the discrepancies between *in vivo* and *in vitro* measurements of the human heel pad properties as the heel pad paradox. Aerts et al. (1995) partially resolved these different results by testing *in vitro* isolated heel pads by the pendulum impact method and by various Instron techniques. They found that heel pad mechanical properties measured were similar using either technique if the tests were carried out in an analogous manner. The pendulum testing needed to be carried out with the heel pad fixed to a solid wall as testing at one site showed that the viscoelastic properties of a wall used as a restraining surface had effected the pendulum results. Both pendulum tests against a rigid wall and Instron tests gave similar results but they were still markedly different from pendulum *in vivo* results. Aerts et al. (1995) concluded that the remaining differences between *in vivo* and *in vitro* results were due to the presence of the lower leg and supporting structures but did not quantify their influence.

Previous studies have not identified the processes in the lower leg that can account for the different mechanical properties measured using the *in vivo* and *in vitro* techniques. In this study the intention is to test the hypothesis that the soft tissue motion of the lower leg is responsible for softening the impact and dissipating energy above which can be accounted for by the heel pad. To examine this hypothesis a computer simulation model of a pendulum impact experiment will be developed which permits partitioning out the effects of the heel pad, the visco-elastic properties of the knee-wall interface, and the relative motion of the bone and overlying soft tissue on the observed mechanical 'heel pad' properties.

METHODS

A two-dimensional model of the shank and heel pad was developed to simulate a pendulum impact experiment. This model was created using DADS (version 9.00 CADSI, Coralville, IA, USA) and solved as a direct dynamics problem. Model parameters are presented in Table 1. The model consisted of an impacting pendulum, a fixed table and wall, a heel pad, and a model element representing the shank (Figure 1).

Table 1 - Model parameters (except for those of the heel pad which are presented in the text)

Model Parameter	Parameter Value
Pendulum Length	2.0 m
Pendulum Mass	11.78 kg
Impact Velocity	1.0 m/s
Mass of Bone	1.70 kg
Moment of Inertia of Bone*	1.80×10^{-2} kg m^2
Mass of Soft Tissue	1.70 kg
Moment of Inertia of Soft Tissue*	2.01×10^{-2} kg m^2
Friction: Static Coefficient	0.3
Friction: Dynamic Coefficient	0.3
Knee–wall Stiffness	5×10^5 N/m
Standard Knee-wall Damping	500 Ns/m
Tendon Stiffness (K_T)	5.78×10^{10} N/m^3
Tendon Damping (C_T)	250 Ns/m

(* moments of inertia are with reference to a transverse axis through the center of mass.)

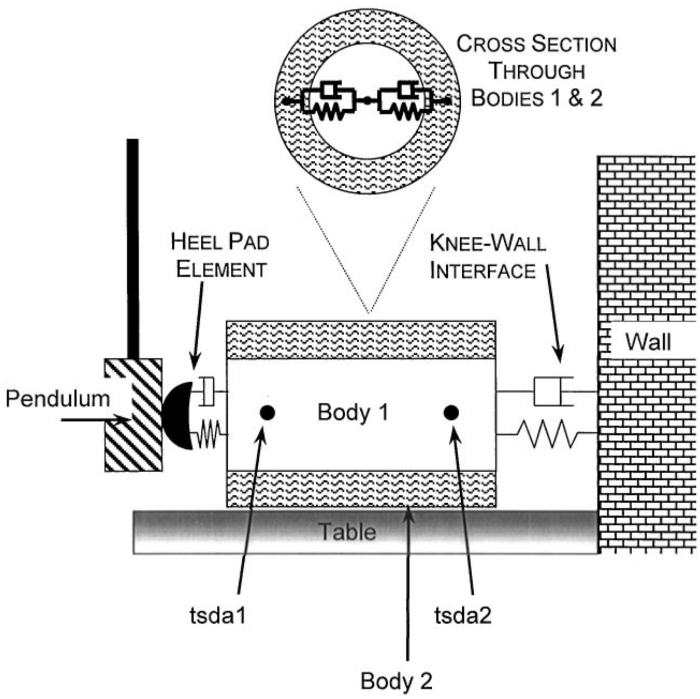

Figure 1 - Diagram of the shank, table, and wall as the pendulum makes impact. Body 1 represents the rigid skeletal structure, and body 2 the wobbling mass; they are connected by two translational spring dampers (see cross section through bodies 1 and 2).

The shank was made up of two bodies connected by non-linear translational spring damper actuators (TSDA). One body represented the bones of the shank and the other represented the soft tissue (potential wobbling mass). Each body was given mass and moment of inertia values comparable with those of a real shank. The mass of a complete shank for a 50th percentile male was taken from Chaffin and Andersson (1991). This mass was then split between the bone mass and the soft tissue mass using the same ratios as Gruber et al. (1998). The two rigid bodies, representing the skeleton and the soft tissue, were connected by two passive TSDA with a free length of zero meters. They were attached at both ends, in 14% of the body's length (see Figure 1). Any independent horizontal or vertical motion of the body's altered the length of the TSDA ($\Delta\mathbf{L}$). The TSDA were given elastic properties similar to those used in Gruber et al. (1998). Damping was chosen so that it near critically damped the system. The TSDA represented the total restoring force ($\mathbf{F}$) between the two bodies not just tendon actions. The equation to represent the TSDA is

$$\mathbf{F} = \mathbf{K_T} \cdot \Delta\mathbf{L}^3 - \mathbf{C_T} \cdot \dot{\mathbf{L}} \qquad [1]$$

where $\mathbf{K_T}$ and $\mathbf{C_T}$ are constants.

The shank was positioned so that it lay on the rigid body of the table and the force components of the knee-wall interface were in contact, to avoid impact chattering. Vertical motion of the bone was allowed and so relative motion of the bone and soft tissue could occur in both directions. Friction was included between the shank and the table by using a friction force element that depended on normal force, user defined coefficients of friction and the contact length. The coefficients of friction were chosen to represent those of skin (Sharkey, 1999).

Contact between the pendulum and the shank was represented by a non-linear contact element designed to represent the heel pad. The heel pad needed to be formulated as a continuous function and as such does not match the instron responses exactly in the transition zone between low and high stiffness. However the key mechanical properties of this element were representative of those measured on isolated heel pads by Aerts et al. (1995). (Table 2)

Table 2 - Key mechanical properties of the heel pad element compared to those measured on isolated heel pads by Aerts et al. (1995)

	% Energy loss	Deformation (mm)	Stiffness (kN/m)*
Aerts et al. (1995)	48	6.3	1080
Model	45	7.1	965

* stiffness measured in the high stiffness zone.

This was placed on the midline of the bone to simulate an impact that produced minimum turning and wobble. The heel pad properties are described by equation 2,

$$\mathbf{F_H} = \mathbf{K_1} \cdot \Delta\mathbf{X}^7 + \mathbf{K_2} \cdot \Delta\mathbf{X}^5 + \mathbf{K_3} \cdot \Delta\mathbf{X}^3 + \mathbf{K_4} \cdot \Delta\mathbf{X} - \mathbf{C} \cdot \dot{\mathbf{X}} \cdot \Delta\mathbf{X} \qquad [2]$$

where $\mathbf{F_H}$ - force acting on the heel pad, $\Delta\mathbf{X}$ - deformation of the heel pad, and the constants have the following values:

$\mathbf{K_1} = 3 \times 10^{15}$ N/m^7
$\mathbf{K_2} = 6 \times 10^{13}$ N/m^5
$\mathbf{K_3} = 3 \times 10^{9}$ N/m^3
$\mathbf{K_4} = 3 \times 10^{4}$ N/m
$\mathbf{C} = 75000$ Ns/m^2

The properties of the heel pad model are illustrated in Figure 2.

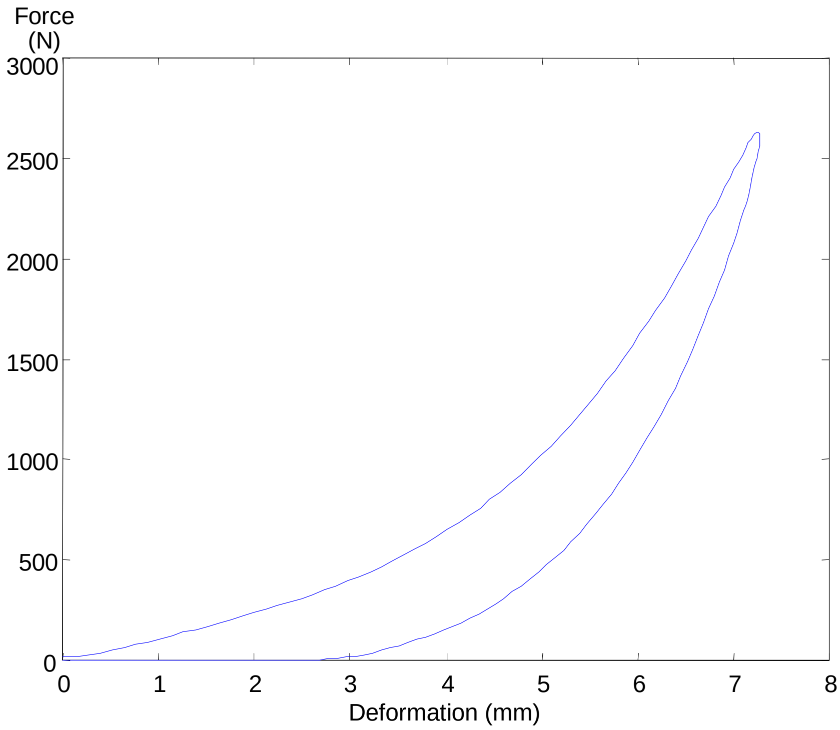

Figure 2 - Force deformation curve of the isolated heel pad model.

A spring-damper element was placed between the knee end of the shank and the wall to represent possible deformations at the knee-wall interface. It represents phenomenologically the properties of the wall and is based on the data of Aerts et al. (1995) and Cavanagh et al. (1984). It has a stiffness value which provides an effective stiffness for the wall at least an order of magnitude greater than the heel pad and allows 2 mm of motion (Cavanagh et al., 1984). The damping value chosen as standard gave energy losses within 1 percent of those found in Aerts et al. (1995).

The model was run in two modes as a wobbling-mass model, and as a rigid body model where the shank was considered to be a single rigid body. The effect of introducing a compliant knee-wall interface, a wobbling mass, and off center pendulum strikes were all examined. Valiant (1984) and Aerts et al. (1995) both stated that off center impacts are problematic as they introduce further energy dissipation effects. Only trials that were visibly not off center were analyzed in these studies. The off center pendulum impacts were changed in 1 mm increments from 0 mm to 10 mm vertically from the center of the heel pad element which was aligned with the long axis of the model element representing the bones of the shank.

Force-deformation curves for the heel pad could be obtained directly by analyzing the contact elements (as in *in vitro* studies), and indirectly by measuring the acceleration-position curve of the pendulum (as in *in vivo* studies). The gradient of the force deformation curves were used to find heel pad stiffness. For the heel pad data the energy lost was computed by integrating the force-deformation curve to find the area in the loop. For the pendulum, energy loss was computed from the velocity of the pendulum before and after impact. These computations mirror those used *in vitro* and *in vivo* to determine heel pad properties, and therefore permit comparison of the mechanical properties of the heel pad as measured by these different techniques.

RESULTS

The heel pad force-deformation curves determined varied depending on the model used and whether they were determined using pendulum measurements or heel pad measurements (Figure 3). A rigid body model gives pendulum measurements and heel pad measurements that are effectively the same. As the pendulum measurements of force require the double differentiation of the displacement data slight differences between the two curves can be seen. Differences due to this process have very little effect on the overall discrepancy between the two measurement techniques (e.g. Figures 3 a and b). Introducing a slightly damped knee-wall interface gives pendulum measurements and heel pad measurements that are clearly different (Figure 3 c and d). If a wobbling mass model is used the inferred heel pad properties from the force-deformation curves, are changed for both measurement techniques (Figures 3 e and f). For the wobbling mass model, deformation characteristics as measured by the pendulum are similar to *in vivo* results reported in the literature. If energy loss and stiffness values are computed, Table 3, the wobbling mass has the greatest influence on producing a discrepancy between the actual properties of the heel pad and those measured by the pendulum. Energy loss is much greater for the wobbling mass model compared with the rigid body model for pendulum based measurements.

Table 3 - Energy loss and stiffness comparisons for pendulum and heel pad measurements for the two models

	Pendulum Measurements		**Heel Pad Measurements**	
	% Energy Loss	Stiffness (kN/m)	% Energy Loss	Stiffness (kN/m)
Rigid Body Model	45.0	935	44.7	962
Rigid Body Model with Knee-Wall Interface	34.4	335	45.0	781
Wobbling Mass Model	89.9	196	45.9	485

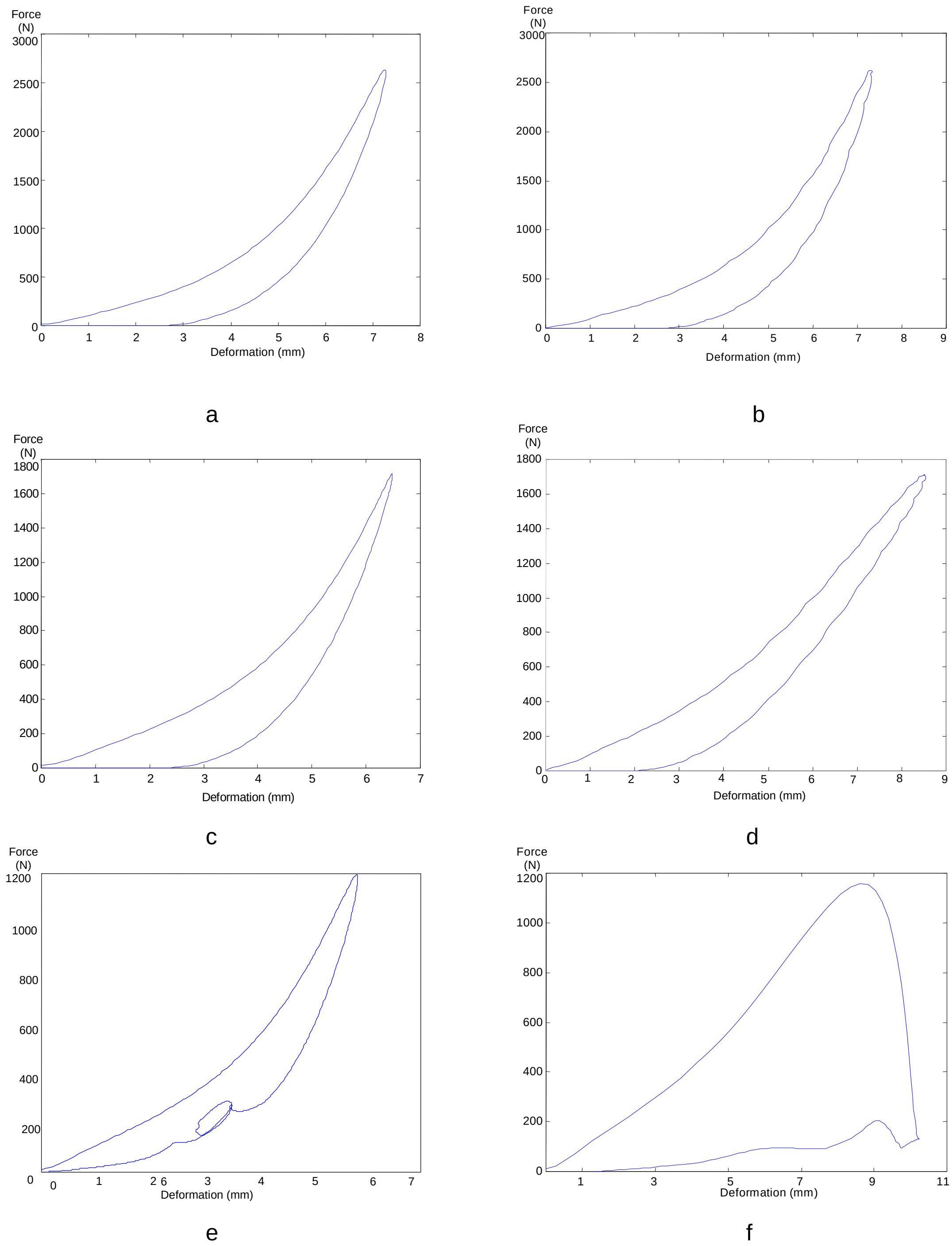

Figure 3 – The heel pad force-deformation curves on the left are those measured directly from the heel pad element and those on the right by the pendulum.
a) and b) the curves for a rigid body model interfaced to a rigid wall.
c) and d) the curves for a rigid body model interfaced to a deformable wall.
e) and f) the curves for a wobbling mass model interfaced to a deformable wall.

Varying the damping of the knee-wall interface from a perfect spring to an overly damped system changed the percentage of energy lost from 29.4% to 51.0% for the rigid body model (Table 4). For the wobbling mass model the energy loss varied from 89.5% to 90.2%. The drop in peak force between the fully rigid model and the final wobbling mass model was 54%. The wobbling mass model reduced the sensitivity of pendulum measures to deformation of the restraining structure, which in experiments is typically a wall.

Table 4 - The effect of softening the knee-wall interface on the energy loss as measured by the pendulum for the rigid body and wobbling mass model

Damping value for Knee-wall Interface (Ns/m)	**Rigid Body Model Energy Loss (%)**	**Wobbling Mass Model Energy Loss (%)**
0	29.4	89.5
500	34.4	89.9
3000	45.2	90.1
10000	51.0	90.2

Varying the impact point between the heel and pendulum had little effect on the rigid body model, for example with the energy lost during an impact (Table 5). The influence of these variations were greater for pendulum measures with the wobbling mass model. Minimal energy loss for the wobbling mass model occurred for a slightly off center strike. For an impact that was off center by more than two millimeters the pendulum lost all of its energy due to the induced rotation of the bone allowing the pendulum to start into its upswing.

Table 5 - The effect of off center impact points on the energy loss as measured by the pendulum for the rigid body and wobbling mass model.

Distance From Center Line of Impact Point (mm)	**Rigid Body Model Energy Loss (%)**	**Wobbling Mass Model Energy Loss (%)**
+10	39.8	-
+2	34.7	99.1
+1	34.4	85.1
0	34.4	89.9
-1	34.7	89.1
-2	34.8	99.2
-10	35.4	-

DISCUSSION

The mechanical characteristics of the human heel pad and shank were investigated using a model. By systematically introducing deformable elements into the model in addition to the heel pad, the heel pad properties determined using a pendulum go from those found experimentally *in vitro* to those found *in vivo* (Figure 3 b and f). Introducing a wobbling mass increased the energy loss during an impact from similar to that found *in vivo*, and reduced the stiffness of the heel pad to that found *in vivo*. These results provide further resolution of the heel pad paradox identified by Aerts et al. (1995). An important implication of these results is that both the heel pad and the soft tissue of the shank make significant contributions to the dissipation of energy during the impacts associated with gait.

While a simplification of the actual system the model appears valid as the model results are comparable to published experimental data. The heel pad stiffness values, determined from the pendulum data, changed from around 1000 kN/m for the rigid model to 200 kN/m for the wobbling mass model with a visco-elastic knee-wall interface. These changes in stiffness are similar in magnitude to those reported for *in vitro* and *in vivo* experiments respectively (e.g. Aerts et al., 1995). The drop in peak force between *in vitro* to *in vivo* experiments was 55 %, (Aerts et al., 1995), within 1% of the value found here.

The deformation of the 'heel pad' increased from 7 mm to nearly 11 mm. During *in vivo* testing of heel pads Kinoshita et al. (1993) and Valiant (1984) reported heel pad deformations of 11.3 mm and 10.4 mm respectively. In all wobbling mass model simulations the maximum excursion of any part of the wobbling mass relative to the underlying rigid structure was less than 17 mm from its original position. Soft tissue motion during running impacts has been found to be over 40 mm (Cappozzo et al., 1996), indicating that it is possible *in vivo* to induce greater soft tissue motion than was examined in the simulation model.

The results also indicate some important considerations when testing heel pads *in vivo*. Off center pendulum impacts had a great effect on the behavior of the wobbling mass model as they induced a relative turning moment and thus produced a large amount of intra-segmental motion (Table 5). Aerts et al. (1995) commented on this phenomenon but did not quantify it. The fact that minimal energy loss did not occur with a dead center hit but with one that was off slightly indicates how sensitive this model is to the pendulum impact initial conditions. The results suggest that for heel impact studies very accurate alignment of the heel pad and pendulum are required, such a finding warrants investigation experimentally. Also given that the ankle is not a simple pin joint it seems unlikely that rotation can be avoided and this could have a noticeable effect on the measured energy loss.

Aerts et al. (1995) identified that a visco-elastic knee-wall interface influenced pendulum determined heel pad properties. The mechanical properties of the knee-wall interface were modeled as a spring-damper that only allowed 2 mm of deformation. With the damping value set at 500 Ns/m the energy loss was similar to that reported for the flexible wall in Aerts et al. (1995), Table 3. There was little difference between the energy loss for the wobbling mass model with a rigid or visco-elastic knee-wall interface (Table 4). This was due to the dominant role of the wobbling mass in energy dissipation. Indicating that the knee-wall interface is not an important variable to consider when testing heel pad and shank properties *in vivo*.

During impacts the heel pad acts to dissipate energy. This study has indicated that the soft tissue of the shank has a similar role. The relative contribution of this wobbling mass has not been quantified before. The role of this wobbling mass is significant, for example peak forces obtained with a heel pad connected to a solid shank were over 100% greater than those for a heel pad connected to a shank with a wobbling mass. The role of the wobbling mass appears to be important and certainly warrants further investigation.

CONCLUSION

The difference between *in vivo* and *in vitro* measured values of heel pad mechanical properties has been described as paradox. A model of the heel pad and lower leg was used to simulate the pendulum experiments used to assess heel pad properties *in vi*vo. Model simulations demonstrated that heel pad properties determined via pendulum experiments actually represent the characteristics of the whole system not just the heel pad. The ability of the wobbling mass of the shank to dissipate energy during an impact was found to be significant. These results demonstrate the important role of both the heel pad and soft tissue of the shank to the dissipation of mechanical energy during impacts.

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