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The influence of soft tissue movement on ground reaction forces, joint torques and joint reaction forces in drop landings

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ABSTRACT

The aim of this study was to determine the effects that soft tissue motion has on ground reaction forces, joint torques and joint reaction forces in drop landings. To this end a four body-segment wobbling mass model was developed to reproduce the vertical ground reaction force curve for the first 100 ms of landing. Particular attention was paid to the passive impact phase, while selecting most model parameters a priori, thus permitting examination of the rigid body assumption on system kinetics. A two-dimensional wobbling mass model was developed in DADS (version 9.00, CADSI) to simulate landing from a drop of 43 cm. Subject specific inertia parameters were calculated for both the rigid links and the wobbling masses. The magnitude and frequency response of the soft tissue of the subject to impulsive loading was measured and used as a criterion for assessing the wobbling mass motion. The model successfully reproduced the vertical ground reaction force for the first 100 ms of the landing with a peak vertical ground reaction force error of 1.2 % and root mean square errors of 5% for the first 15 ms and 12% for the first 40 ms. The resultant joint forces and torques were lower for the wobbling mass model compared with a rigid body model, up to nearly 50% lower, indicating the important contribution of the wobbling masses on reducing system loading.

Key words: wobbling mass, soft tissue, joint torque, forward dynamics, landings

INTRODUCTION

To try and circumvent the many problems associated with internal force measurement in man, inverse dynamics and computer modeling are commonly used. Biomechanical whole body models are normally composed of rigid segments linked by simple kinematic connections (e.g. Bobbert and van Soest, 1994; Gerritsen et al., 1996). However, the segments of the human body are not rigid and such an assumption can lead to substantial errors in both inverse and direct dynamics analyses, especially those associated with high accelerations and impulsive loading. These types of activity are often associated with injuries or discomfort (Nigg and Bobbert, 1990). As modeling these activities is one of the few methods of obtaining joint loading information it may be very important that the model can account for soft tissue motion and the kinetic effects it has on the body.

Models which accommodate some force interactions within a body segment have received limited attention (Minetti and Belli, 1994, Cole et al., 1995; Gruber et al., 1998; Wright et al., 1998; Nigg and Liu, 1999, and Liu and Nigg, 2000). Typically in these models segments are separated into two elements: a rigid component, and a soft tissue component - the wobbling mass. Minetti and Belli (1994) and Wright et al. (1998) only included a single wobbling mass to represent the visceral mass. Nigg and Liu (1999) and Liu and Nigg's (2000) model of the impact phase of running considered vertical motion only. The model represented the body as two segments (upper and lower body) each consisting of rigid and wobbling masses connected with springs and dampers.

Gruber et al. (1998) used a two-dimensional, three segment wobbling mass model to recreate the vertical ground reaction force for a subject landing from a drop. They showed that ground reaction forces and joint torques and forces were markedly different for their wobbling mass model and an equivalent rigid body model when simulating the same landing from a drop. However, a large number of model parameters, including mass distributions between segment rigid and wobbling mass elements were optimized to achieve a ground reaction force match between model and experimental data. The distributions of segmental mass between skeletal and soft tissue components were well beyond the ranges indicated from dissection (e.g. Clarys et al., 1984). Model joint torques were zero until five milliseconds after impact, but inverse dynamics analysis of landings show significant joint torques prior to impact (Bobbert et al., 1992). Pain and Challis (2004) demonstrated the sensitivity of such wobbling mass models to their model parameters and showed that compensating errors could account for anomalies such as these.

Cole et al. (1996) produced a two-dimensional, four segment wobbling mass model to examine joint loading during impact in running. In this model the mass of the bone and soft tissue were calculated from the tissue distributions in Clarys and Marfell-Jones (1986). The soft tissue elements were point masses constricted to move along the line of action of the muscle-tendon tendon unit and had a moment of inertia of zero. The soft issue motion being restricted to one line of action and having no moment of inertia would greatly reduce the kinetic contributions of this element. As the soft tissue would be the dominant contributor to the inertial properties of the segment, and as soft tissue motions have been recorded in all three planes (Reinchmidt, 1996), these assumptions may limit this model's ability to examine soft tissue motion affects.

Previous studies have shown the potential influence of the wobbling masses on system kinetics, but these studies have suffered from a variety of deficiencies. These deficiencies include unrealistic model parameters, constrained wobbling mass motion, and joint torque patterns which are not observed experimentally. The aim of this study was to determine the effects that soft tissue motion have on ground reaction forces, joint torques and joint reaction forces in drop landings. To this end a four body-segment wobbling mass model was developed to reproduce the vertical ground reaction force curve for the first 100 ms of landing. Particular attention was paid to the passive impact phase, (occurring in the first 50 ms, Nigg, 1986), while selecting most model parameters a priori, thus permitting examination of the rigid body assumption on system kinetics.

METHODS

Measurements were performed on an experimental subject performing drop landings, and a model was developed to simulate these landings.

The subject was a male, age 27 years, height 1.75 m, mass 85 kg, body fat 10% of total body mass, who had provided informed consent. The subject performed two two-footed landings from a drop height of 0.43 m, making initial contact with the heels. The subject had reflective markers on the lateral second metatarsal, the lateral malleolus, the heel, the center of rotation of the knee, the greater trochanter, and the shoulder. The drops were performed barefooted and the arms were squeezed tight across the chest to minimize arm motion. Force plate data (Bertec, N50601, Type 4080s) were recorded at 1200 Hz, and the marker motion data were recorded at 240 Hz (Pro-Reflex, Qualisys, Sweden). During these landings segment orientations at impact differed by less than one degree, and peak ground reaction forces by less than 6 percent. Therefore initial segment orientations for the model at contact were the mean data from these two trials.

It was not feasible to measure soft tissue motion during the landings, therefore to obtain representative data soft tissue motion was measured during controlled impacts, using the methods presented in Pain and Challis (2002). The subject was positioned so that he could strike a force plate with a vertical downward stamping motion with the knee flexed at 90^{o} and that allowed the motion of an array of 28 markers on the posterior aspect of the shank to be recorded at 240 Hz, the shank test. The stamping motion was performed such that a rigid beam with a padded surface provided support for the thigh at impact. Six trials were performed. The process was repeated with the marker array on the anterior aspect of the left thigh with the leg straight, the thigh test, and the upper body and other leg were supported at impact. From these data mean marker array motion was determined, and the magnitude and frequency content of the experimental soft tissue motion computed for the passive impact phase for later comparisons with the model.

A two-dimensional model, Figure 1, consisting of four rigid links (bone) connected with revolute joints, controlled by revolute spring-damper actuators, had wobbling masses (soft tissue) attached to the shank, thigh and torso bones with translational spring-dampers (Pain and Challis, 2001). The ground-heel interface was represented by a non-linear spring-damper system described in Pain and Challis (2001). The model was developed in DADS (version 9.00, CADSI) to simulate landing from a drop of 43 cm. Simulations could also be run with the model as a rigid body model by fixing together the centers of mass of the bone and the soft tissue for each body segment using rigid joints.

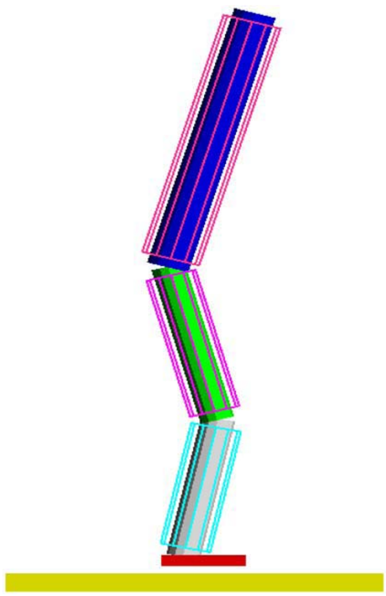

Fig. 1. Schematic of the four body-segment wobbling mass model just before impact. Inner solid segments represent the rigid skeleton. The outer line segments represent the wobbling mass material.

Body segment inertial parameters for the subject's segment mass and location of the center of mass were calculated using the equations of Zatsiorsky et al. (1990). Lower limb segment moments of inertia were calculated using the equations of Challis (1996). The mass of the shank and thigh were divided for each segment into a bone mass and soft tissue mass by modeling them as a cylinder and a tube respectively. The cadaver data of Clarys and Marfell-Jones (1986) and Clarys et al. (1984) were used for the relative masses and density of the rigid and wobbling components. The radii of the cylinder and the tube were systematically adjusted so that the total mass and moment of inertia of the two geometric solids corresponded to the subject's anthropometry. The upper body was modeled as one body composed of rigid and wobbling mass components. The mass distributions of these segments was based on the data of Clarys et al. (1984) and Clarys and Marfell-Jones (1986). The rigid component was modeled as a series of cylinders, each with different densities, representing the pelvis, spinal column, and the head to calculate the moments of inertia of the trunk skeleton. The soft tissue of the trunk was modeled as a tube surrounding the bone of the trunk, and the arms were modeled as a cylinder held across the chest to represent the arms crossed in front of the chest.

In free fall immediately before contact soft tissue motion is minimal, as indicated by the invariant area of four additional markers placed on the bellies of the muscles of the thigh and shank. However, during impacts segment motion determined by bone and soft tissue mounted markers can differ by up to 10^{o} (Reinschmidt, 1996). Due to the paradoxical problem of determining bone orientation from surface markers during impacts the parameters for the torque generators were determined so that both the bone and soft tissue segments of the model were within one degree of the subjects body segment angles at 40 ms after impact.

All angles were measured with reference to the vertical, with clockwise rotations positive. On the subject angles were defined with respect to the vertical and the line joining the joint centers. In the model angles were defined with respect to the vertical and the midline of the bone or soft tissue segments. These model values were then used in the final version of the wobbling mass model. During this phase of model parameter identification the stiffness and damping of the spring-dampers, which connect the soft tissues to the rigid body, were constrained so that the bone and soft tissue segments remained within one degree of each other. The torque generators at the joints, the revolute spring-dampers, provided joint torques at the instant of impact.

The final model adjustments were the stiffness and damping of the spring-dampers, which effectively connect the soft tissues to the rigid body. The tendon properties were altered by up to one order of magnitude (Pain and Challis, 2004) to produce a vertical ground reaction force that matched the subject's. The veracity of these changes was assessed by comparing the motion of the wobbling masses in the model to the measured soft tissue motion on the subject during the shank and thigh impact experiments.

RESULTS

The model parameters and initial conditions are presented in the tables 1, 2 and 3. Table 1 presents the subject specific bone and soft tissue inertial parameters. The subject's body segment angles at impact and 40 ms after impact are presented in Table 2, the model had the same segment angles at impact and attempted to reproduce the same joint angles 40 ms after impact. The model had variable joint torques throughout the impacts these were produced by rotational spring-damper actuators at the joints, their model parameters are described in Table 3.

Table 1. The inertia parameters for the bone and soft tissue calculated for the subject

Body segment	Segment type	Mass (kg)	Moment of inertia (kgm^2)	Segment length (cm)	Center of mass above midpoint (cm)
Foot	Whole	2.20	3.42×10^{-3}	26.5	0.60
Shank	Bone	2.68	3.80×10^{-3}	41.0	2.75
	Soft tissue	5.56	0.0132	41.0	2.75
Thigh	Bone	4.30	0.0570	42.5	2.85
	Soft tissue	13.42	0.240	42.5	2.85
Trunk	Bone	6.20	0.447	86.3	10.0
	Soft tissue	53.40	1.44	86.3	10.0

Table 2. Subject body segment orientations at impact and 40 ms after impact for the two drop landings

	Foot	Shank	Thigh	Trunk
Initial angle (deg)	-92	-175	165	-165
Angle at 40 ms (deg)	-88	-172	147	-149

Table 3. Model joint rotational stiffness and damping model coefficients

	Ankle	Knee	Hip
Stiffness (N / ^{o})	70	10	15
Damping (N.s / ^{o})	0.50	0.35	0.30

Soft tissue motion was measured during controlled impacts by the subject, and was quantified by the magnitude of marker motion and frequency content of that motion. The experimental and simulation marker motions and frequency contents compared very favorably (Table 4). For the six trials of shank test the mean peak vertical ground reaction force was 4615 $\pm$ 340 N, and for the thigh test 6113 $\pm$ 502 N, these forces were for one leg only. For the experimental two footed impacts the peak vertical ground reaction force was 13675 N.

Table 4. Comparison of magnitude and frequency content of experimental soft tissue motion and model soft tissue motion

	Shank		Thigh	
	Magnitude (cm)	Peak Frequencies (Hz)	Magnitude (cm)	Peak Frequencies(Hz)
Experimental	1.8 $\pm$ 0.2	14, 28, 50	3.2 $\pm$ 0.9	14, 18
Model	1.4	12, 24, 39	2.8	15, 20

N.B. - For experimental values this is the mean marker motion across markers. For the model it is the relative motion of the center of mass of the bone and soft tissue for each body-segment.

Given the model parameters the first 100 ms of an impact from a 0.43 m drop were simulated. The model reproduced the experimental vertical GRF for the first 15 ms of the landing within 5% and the first 40 ms within 12% (Figure 2). Between 40 ms and 80 ms it reproduced the shape of the curve well and key values such as the descending shoulder were close to experimental values. The peak GRF for the wobbling mass model was 16.2 bodyweights, and for the subject 16.4 bodyweights. During these simulations the maximum difference in orientation between the bone segment and the soft tissue segment in the model was up to one degree in the shank, and 4.5^o in the thigh. For example the orientations of the bone segments were re-examined at 40 ms after impact for the wobbling mass model, the angles were -91^o, -171^o, 150^o, and -151^o for the foot, shank, thigh, and trunk respectively. These values correspond well with the experimental data (Table 2). The difference between the models trunk orientation and the experimental data was no greater than 2^o throughout the simulated motion.

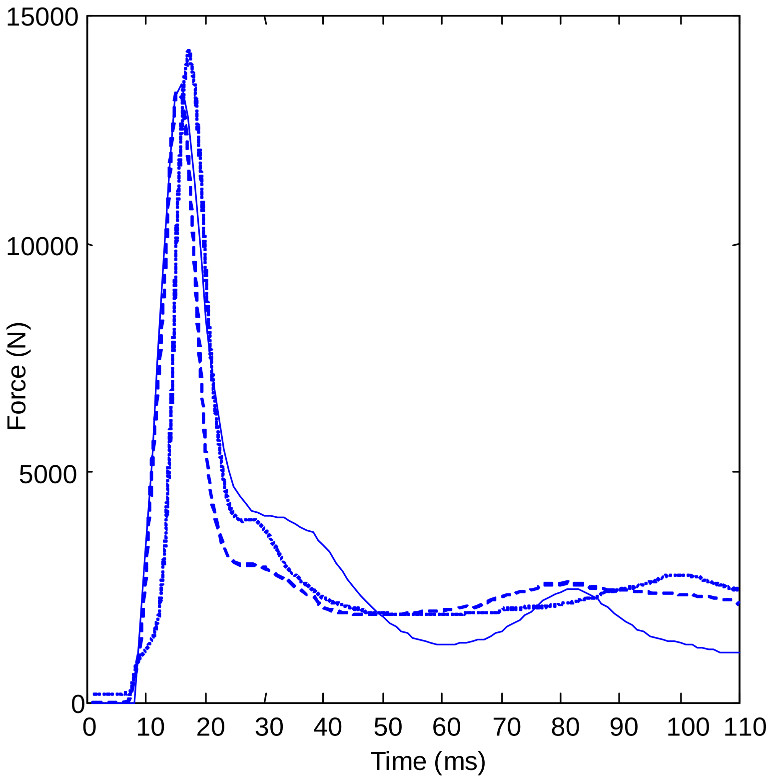

Figure 2. Vertical ground reaction force curves for the two empirical trials (dotted line and dashed line) and the wobbling mass model (solid line).

Peak joint torques and forces were much greater for the rigid body model compared with the wobbling mass model (Table 5). The orientation of the bone segments differed by less than 2^o between the wobbling mass and the rigid body models. With a fully rigid model the peak vertical ground reaction force increased to 40.5 bodyweights, compared with 16.2 bodyweights for the wobbling mass model. With rigid legs, and only a wobbling mass for the trunk, similar to Wright et al. (1998), the peak vertical ground reaction force was 31.4 bodyweights.

Table 5. Comparisons between the peak joint torques and forces for the wobbling mass and rigid body models

	Wobbling Mass Model		Rigid Body Model	
Joint	Torque (Nm)	Vertical Force (N)	Torque (Nm)	Vertical Force **(N)**
Ankle	-228	11080	-370	17140
Knee	267	7720	500	13280
Hip	-240	5100	-460	7700

DISCUSSION

The aim of this study was to develop a four body-segment wobbling mass model to simulate landing from a drop, so that the influence of the rigid body assumption on system kinetics could be examined. To produce the model as many model parameters as possible were determined prior to the simulations, specifically

- The segment inertial parameters were calculated to match the subject.
- Partitioning of segment mass to rigid and wobbling mass components was based on cadaver data (Clarys et al., 1984; Clarys and Marfell-Jones, 1986).
- A heel pad model was adopted (Pain and Challis 2001).
- Initial configuration and velocity of the model at impact were determined from subject kinematics.

The remaining model parameters were those for the rotational spring-damper actuators, and the stiffness, and damping of the spring-dampers connecting the rigid and wobbling masses. An independent test of soft tissue motion compared very favorably with the model produced motion (Table 4), providing a level of confidence in the model parameters.

The model was successful in reproducing the vertical ground reaction force curve for the passive impact period, the first 40 ms. The overall shape of the curve matches well up to 100 ms, and the force values of the peak and descending shoulder are very similar. With a rigid body model the system kinematics were similar to the experimental subject's; the differences were within the anticipated experimental error in measurements of the segment orientations. Despite these similarities in the kinematics the peak vertical ground reaction forces were 16.4, 16.2, and 40.5 bodyweights, for the subject, wobbling mass model, and rigid body model. Similarly resultant joint moments were much greater for the rigid body model compared with the wobbling mass model (Table 5). These results provide evidence of the important role of soft tissue motion in reducing joints loads for this task. The task selected here parallels that used by Gruber et al. (1998), and reflects a condition which can occur during landings from a jump, especially somersaults, and provides links to a running where most impacts are via the heel. Further studies should examine if these phenomena exist for other activities involving impacts, for example walking and running.

Comparing the joint torques and forces between the wobbling mass model and a rigid body model show the same overall trend as in Gruber et al. (1998). The wobbling mass model decreased forces and torques at the joints. However here the change was not as drastic as seen in Gruber et al. (1998) where hip torques varied by almost an order of magnitude. The results in Gruber et al. (1998) can be attributed to erroneous timing of the activation of the torque actuators in their wobbling mass model.

Sensitivity analyses of the results demonstrate that the trunk wobbling mass was almost solely responsible for the peak in the vertical ground reaction force at 80 ms. The model was weakest in its representation of the trunk, as there was no impact data to separately determine the magnitude and frequency response of the torso as there was for the shank and thigh. It is feasible that the viscera and musculature of the trunk are acting over different time scales. Unfortunately no information is available on the response of the viscera to an impulsive load. Minetti and Belli (1994) measured visceral motion but this was for a forced oscillation and the period of oscillation of the viscera was 5 Hz. The trunk soft tissue mass is undoubtedly a major contributor to the ground reaction force in latter stages of the impact. However, its contribution in this study was not so great that it could be justified to have a model which only had one wobbling mass element which is associated with the trunk segment, for example as used in Wright et al. (1998).

Using experimental segment orientation data for model initial conditions and evaluation is paradoxical. The soft tissue motion obscures the bone motion and accurate measurements of bone motion are not practicable. However, in the free fall phase of the drop this motion is minimal, providing confidence in the experimental data used for the initial conditions. Segment orientations 40 ms into the landing were compared between the model and subject, and were within 3^o. These angles were obtained from surface marker data and so were influenced by soft tissue motion. The bone segment orientations and the subject body segment orientations may have been up to 3^o different at 40 ms but the corresponding soft tissue positions and orientations gave possible orientations for the composite segment that were equal to the subject body segment angles. Without measured subject bone angles and whole limb angles a true comparison of orientations is not possible. However these angles were not used as model inputs but were used to examine if the model was giving similar kinematics to the experimental subject, which they appear to do well within the bounds of the indeterminate nature of the problem.

CONCLUSION

A wobbling mass model was developed using subject specific parameters. Most of the parameters were limited within ranges obtained independently of the landing that was being modeling. With these model parameters the model produced segmental kinematics, ground reaction forces, and soft tissue motion similar to that of an experimental subject. It successfully reproduced the vertical ground reaction force for the first 100 ms of the landing. Although not perfectly matching all aspects of the subject vertical ground reaction force the main discrepancy indicates that a more complex trunk model is necessary. The joint torques and forces calculated in this model were lower than in a rigid body model, and indicate the important role of soft tissue motion during impacts.

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