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Technical note

A mechanical model to determine the influence of masses and mass distribution on the impact force during running

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Abstract

Simple spring–damper–mass models have been widely used to simulate human locomotion. However, most previous models have not accounted for the effect of non-rigid masses (wobbling masses) on impact forces. A simple mechanical model of the human body developed in this study included the upper and lower bodies with each part represented by a rigid and a wobbling mass. Spring–damper units connected different masses to represent the stiffness and damping between the upper and lower bodies, and between the rigid and wobbling masses. The simulated impact forces were comparable to experimentally measured impact forces. Trends in changes of the impact forces due to changes in touch-down velocity reported in previous studies could be reproduced with the model. Simulated results showed that the impact force peaks increased with increasing rigid or wobbling masses of the lower body. The ratio of mass distribution between the rigid and wobbling mass in the lower body was also shown to affect the impact force peak, for example, the impact force peak increased with increasing rigid contribution. The variation in the masses of upper body was shown to have a minimum effect on the impact force peak, but a great effect on the active force peak (the second peak in the ground reaction force). Future studies on the dynamics and neuro-muscular control of human running are required to take into consideration the influence of individual variation in lower body masses and mass distribution.

Keywords: Impact; Force; Running; Mass; Human; Model; Dynamics

1. Introduction

Vertical ground reaction forces in heel–toe running have a high-frequency impact force peak during the first 50 ms of contact (Nigg, 1978; Cavanagh and Lafortune, 1980; Frederick et al., 1981). Special attention has been paid to midsole cushioning of running shoes in order to reduce the impact force peak and running injuries (Andreasson and Peterson, 1986). However, the effect of variation of midsole hardness of running shoes on vertical impact force peaks has been surprising: they did not differ significantly when running with shoes of varying midsole hardness (Clarke et al., 1983; Nigg et al., 1983,1987; Snel et al., 1985). The mechanism behind the observations has not been understood clearly and there is a lack of conclusive evidence due to many speculations (Nigg et al., 1987; Bates et al., 1986; Wright et al., 1998). Human body during the impact can be considered as a mechanical system with masses and connecting springs and dampers that changes accordingly to the neuro-muscular control. An examination of the effect of body masses on the impact force is important in understanding the dynamics of human running. This effect, however, is difficult to identify experimentally since it couples with the influence of neuro-muscular adaptation of human body (Bates et al., 1986; Simpson et al., 1989). A modeling study is a useful adjunct to experimental studies, and may provide an insight view of the influence of body masses through simulations with unchanged springs and dampers of the system.

Simple spring–damper–mass models have been successfully used to simulate human running and hopping (Alexander, 1988; Cavagna et al., 1988; Thompson and Raibert, 1989; Farley and Gonzalez, 1996; Ferris and Farley, 1997; Ito et al., 1983; McMahon and Cheng, 1990; Kim et al., 1994; Nigg and Anton, 1994). However, these models assumed only rigid body segments in their simulations. A model of the human body using rigid segments

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is only justified to study slow quasi-static movements and is particularly not appropriate when studying impact situations (Denoth et al., 1984; Gruber et al., 1987; Cole, 1995). Human body corresponds to a mechanical system of rigid (bones) and non-rigid (muscles and other soft tissues) masses, which are attached to each other through elastic and viscous connections. The non-rigid mass has been named *wobbling mass* in a previous publication (Gruber et al., 1987). Varied mechanical components of the system affect the resulting impact loading. It is, therefore, of interest to study the effect of changes in rigid and wobbling masses of the system on the impact forces.

The objectives of this study were to develop a simple spring–damper–mass model simulating human running with rigid mass representing bones and wobbling mass representing soft tissues; and to study the effect of changes in masses and mass distribution on the impact force during simulated heel–toe running assuming an unchanged elastic and viscous properties of the system.

2. Methods

A simplified spring–damper–mass model was used in this study (Fig. 1). It consisted of four masses. The upper body was modeled using two masses, one representing its rigid mass, m_3, and the other representing its wobbling mass, m_4. The thigh, leg and foot of the supporting leg were modeled using two masses, one representing its rigid mass, m_1, and the other representing its wobbling mass, m_2. The total body mass was assumed to be 75 kg. The rigid and wobbling mass parameters used in this study were adapted from Cole (1995) with the modification that the lumped rigid and wobbling masses of the supporting leg (Table 1) were the sum of all the corresponding rigid and soft tissue masses, respectively.

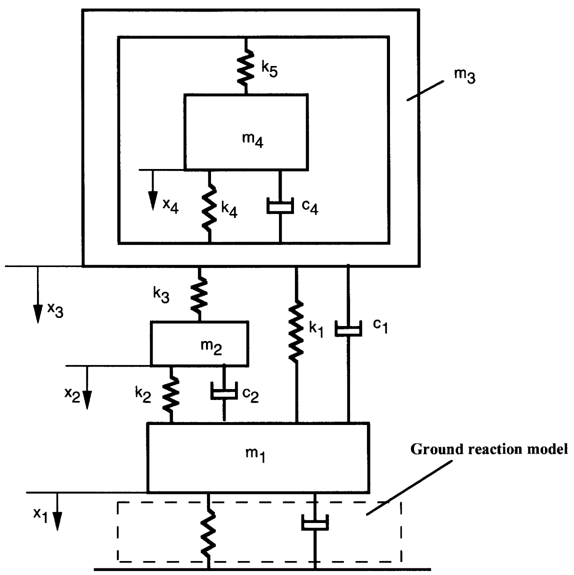

Fig. 1. A simplified spring–damper–mass model used in the current study. Elements of the system shown in the figure are: lower body rigid mass (m_1) and wobbling mass (m_2), upper body rigid mass (m_3) and wobbling mass (m_4), compressive spring (k_1) and damper (c_1) that connect the upper and lower rigid bodies, spring (k_3) and spring–damper unit (k_2, c_2) connecting the lower wobbling mass to the upper and lower rigid bodies, spring (k_5) and spring–damper unit (k_4, c_4) connecting the upper wobbling mass to the upper rigid mass.

Table 1
The parameters (masses, spring constants, and damping coefficients) of the system

M_1 (kg)	M_2 (kg)	M_3 (kg)	M_4 (kg)	K_1 (kN/m)	K_2 (kN/m)
6.15	6	12.58	50.34	6	6
K_3 (kN/m)	K_4 (kN/m)	K_5 (kg/s)	c_1 (kg/s)	c_2 (kg/s)	c_4 (kg/s)
10	10	18	300	650	1900

Spring–damper systems were used to connect the four masses of the model (Fig. 1 and Table 1). The wobbling mass of upper body, m_4, was attached to the rigid mass of the upper body, m_3, through a spring, k_5, at the upper end and a spring–damper combination, k_4 and c_4, at the lower end of the wobbling mass. The values of the spring constants were estimated from muscle–tendon properties present ed earlier (Cole, 1995). The damping coefficient, c_4, was estimated based on the critical damping assumption of muscle–tendon units presented earlier (Hörler, 1972). The wobbling mass of the supporting leg was connected to the upper rigid body through a spring, k_3, and to the lower rigid body through a spring, k_2, and a damper, c_2. The connection between the upper rigid body m_3 and the lower rigid body m_1 included a compressive spring, k_1, and a damper, c_1, similar to the leg spring that was used to represent the overall stiffness and damping properties of the human leg (McMahon and Cheng, 1990; Farley and Gonzalez, 1996).

The ground reaction force model used in this study was a nonlinear visco-elastic force model with combined material properties of heel pad, shoe, and contact surface, similar to the one used by Cole (1995). The vertical contact force, F_g, acting on the foot from ground was determined using

$$F_g = A_c[ax_1^b + cx_1^d v_1^e] \quad (x_1 > 0),$$
$$F_g = 0 \quad (x_1 \leqslant 0).$$

The contact force, F_g, was simulated as a nonlinear function of the deformation (x_1) determined by parameters a and b, and the deformation velocity (v_1) of the contact element determined by parameters c, d and e. The parameters a, b, c, d, and e were assumed to be shoe-specific. Two types of shoe–foot models (soft and hard) were used in the simulations with values of parameters

Table 2
The parameters used for ground reaction models of two pairs of shoes with different midsole hardness. The parameters were determined using a trial and error procedure to fit simulated pendulum impact tests to force–deformation curves determined in pendulum impact experiments on two pairs of experimental running shoes with different midsole hardness (Aerts and De Clercq, 1993)

	a	b	c	d	e
Soft shoe	1.0×10^6	1.56	2.0×10^4	0.73	1.0
Hard shoe	1.0×10^6	1.38	2.0×10^4	0.75	1.0

determined by simulated pendulum impact tests. The force–deformation curves predicted from simulated pendulum impact tests were compared to the force–deformation curves determined in pendulum impact tests on two pairs of experimental running shoes with different midsole hardness (Aerts and De Clercq, 1993). The differences were minimized using a trial and error procedure to find a set of parameters with a best fit between two sets of curves for each pair of shoes (Table 2). The constant A_c was taken to be 2, assuming that the average contact area during ground contact was twice the area of contact between the impact pendulum and the foot in the experiments by Aerts and De Clercq (1993).

The equations of motion for the system illustrated in Fig. 1 are

$$\begin{aligned}
m_1\ddot{x}_1 &= m_1 g - F_g - k_1(x_1 - x_3) - k_2(x_1 - x_2)\\
&\quad - c_1(\dot{x}_1 - \dot{x}_3) - c_2(\dot{x}_1 - \dot{x}_2),\\
m_2\ddot{x}_2 &= m_2 g + k_2(x_1 - x_2) - k_3(x_2 - x_3)\\
&\quad + c_2(\dot{x}_1 - \dot{x}_2),\\
m_3\ddot{x}_3 &= m_3 g + k_1(x_1 - x_3) + k_3(x_2 - x_3)\\
&\quad - (k_4 + k_5)(x_3 - x_4) + c_1(\dot{x}_1 - \dot{x}_3)\\
&\quad - c_4(\dot{x}_3 - \dot{x}_4),\\
m_4\ddot{x}_4 &= m_4 g + (k_4 + k_5)(x_3 - x_4) + c_4(\dot{x}_3 - \dot{x}_4),
\end{aligned}$$

where g is the gravity constant and other elements are described above.

Numerical simulation was executed on an Indigo2 R10,000 station (Silicon Graphics Inc.) using a multibody dynamic simulation software (DADS, version 8.0, CADSI, Oakdale, Iowa). The ground reaction model was implemented in FORTRAN and linked to the DADS simulation program using the 'USERFORCE' option. The vertical movement was simulated for the first 250 ms after touchdown. Simulations did not include take-off phase of running since the model included only passive elements.

The same vertical velocities (0.6 m/s) at touchdown were chosen for all body masses (Aerts and De Clercq, 1993). To validate the model, the effect of changes in the touchdown velocity of the foot ranging from 0.2 to 1.0 m/s with 0.2 m/s increments was examined through simulations. Two different shoe models (hard and soft) were compared in simulations for the same purpose. The effect of changes of the lower rigid body mass, m_1, was evaluated from 2.15 to 10.15 kg with 2 kg increments. The effect of changes of the lower wobbling mass, m_2, was evaluated from 2 to 10 kg with 2 kg increments. The effect of changes in the mass ratio, r_{lm}, between lower rigid mass and lower wobbling mass (with the total mass of the lower body remaining the same) was evaluated with various values of 0.22, 0.52, 1.03, 2.04, and 5.08. The effect of upper masses and mass radio were evaluated by varying m_3 from 6.58 to 18.58 kg with 3 kg increments, varying m_5 from 38.34 to 62.34 kg with 6 kg increments, and varying r_{um} at 0.12, 0.25, 0.42 and 0.64, respectively. The mass ration, r_{um}, was defined as the ratio of upper rigid mass versus upper wobbling mass, and varied while the total mass of the upper body remained the same.

3. Results

Simulated impact force peak was 1353 N for hard shoe corresponding to about 2 times of body weight (Fig. 2a). The duration from first contact to the first peak (impact force peak) was about 23 ms. The hard shoe generated greater impact force peak and higher loading rate than the soft shoe. The loading portion of vertical ground reaction force versus vertical displacement of the center of mass of the upper body showed two different regions with different stiffness (Fig. 2b). An increase in touchdown velocity of the lower body (rigid and wobbling masses) produced an increase (up to 222 N) of the impact force peak in the hard shoe condition (Fig. 2c).

The results of simulations for the hard shoe condition showed that increases in the lower rigid mass produced incremental increases in the impact force peak ranging from 161 to 172 N (Fig. 3a). Increases of the lower wobbling mass increased the impact force peak incrementally up to 86 N (Fig. 3b). Increases in the mass ratio of lower body, r_{lm}, led to incremental increases in the impact force peak up to 126 N (Fig. 3c).

Increases in the upper rigid mass had small effect on the impact force peak with an incremental increase up to 5 N (Fig. 3d). The impact force peak was not affected by the changes of the upper wobbling mass (Fig. 3e). There was a small incremental change up to 8 N in the impact force peak corresponding to the changes of mass ratio of the upper body (r_{um}) (Fig. 3f). Increases in the upper rigid and wobbling masses led to increases in the active force peak, i.e. the second peak in the vertical ground reaction force. The mass radio of upper body had only small effect on the active force peak.

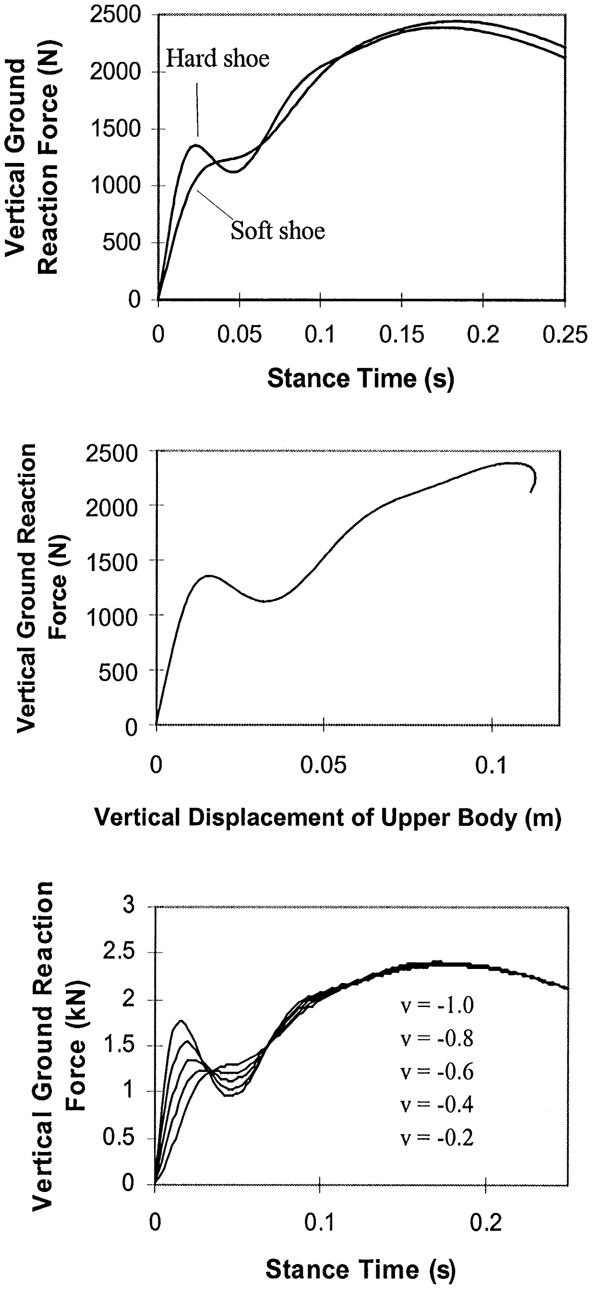

Fig. 2. Simulated results that show (a) the comparison of ground reaction force–time history curves in simulated running with soft and hard shoes, (b) the simulated ground reaction force–deformation of the center of the upper rigid body with hard shoe, and (c) a higher touch-down velocity resulting in a greater impact force peak and active force peak. The values of velocities (m/s) are listed in the figure from bottom up as related impact force peaks increase.

4. Discussion

A simple spring–damper–mass model developed for the study of human running predicted the impact force peaks (2 times of body weight within first 23 ms after touchdown) of running with their values within physiological range of measured values from human running experiments (Nigg, 1986). The simulated results showed that a higher touch–down velocities resulted in a higher impact force peak. This result is in agreement with experimental data (Nigg et al., 1987) and results from other simulation studies (Gerritsen et al., 1995). The simulated characteristics of the vertical ground reaction force versus the vertical displacement of the center of mass of the upper rigid body showed two sets of stiffness that correspond qualitatively to the curves from running experiments (Farley and Gonzalez, 1996).

A one-dimensional model with lumped mass, spring, and damping parameters used in the present study could only approximately predict some principles of the human running. There were limitations for the model and simulations. For instance, constant values of spring and damper parameters did not account for a rapid change in joint angles and muscle activity after the impact phase (Bobbert et al., 1992). Furthermore, unaltered spring and damper parameters for various body masses could not account for a possible neuro-muscular regulation by adjusting the stiffness and damping of the system. The results of simulations, in which hard shoe generated higher impact force peak than the soft shoe, was in accordance with observations in pendulum tests (Aerts and De Clercq, 1993; Lafortune et al., 1996), but not in running experiments (Clarke et al., 1983; Nigg et al., 1983; Snel et al., 1985). On the other hand, it was the intention of the current study to examine only the effect of body masses without altering the spring and damper of the system. Another limitation of the current model was its fixed contact area during ground contact. A prolonged contact period (estimated 380 ms) in the simulated results comparing to 210–340 ms in running experiments (Farley and Gonzalez, 1996) might be the results of a combination of several limitations of the model.

The models with human body being divided into an upper and a lower part have been used in the study of mechanical properties of the paw pads of mammals (Alexander et al., 1986) and the study of impact forces during human jumping (Ozguven and Berme, 1988). However, Gruber et al. (1987) in a modeling study found a substantial difference in the predicted impact forces in the knee and hip joint between the models with and without wobbling mass. Cole (1995) recently showed that the model with rigid and soft tissue masses predicted the ground reaction force peak better than the model with only rigid mass. The upper and lower bodies were further divided into rigid and wobbling masses in the current model. The simulations of the current model showed a strong influence of rigid or wobbling mass in the lower body on the impact force peak without neuro-muscular adaptation. Bates et al. (1986) attached an additional mass to runner's leg similar to an increase in the wobbling mass of the lower body in the current study. They observed a Newtonian response where the effect of the

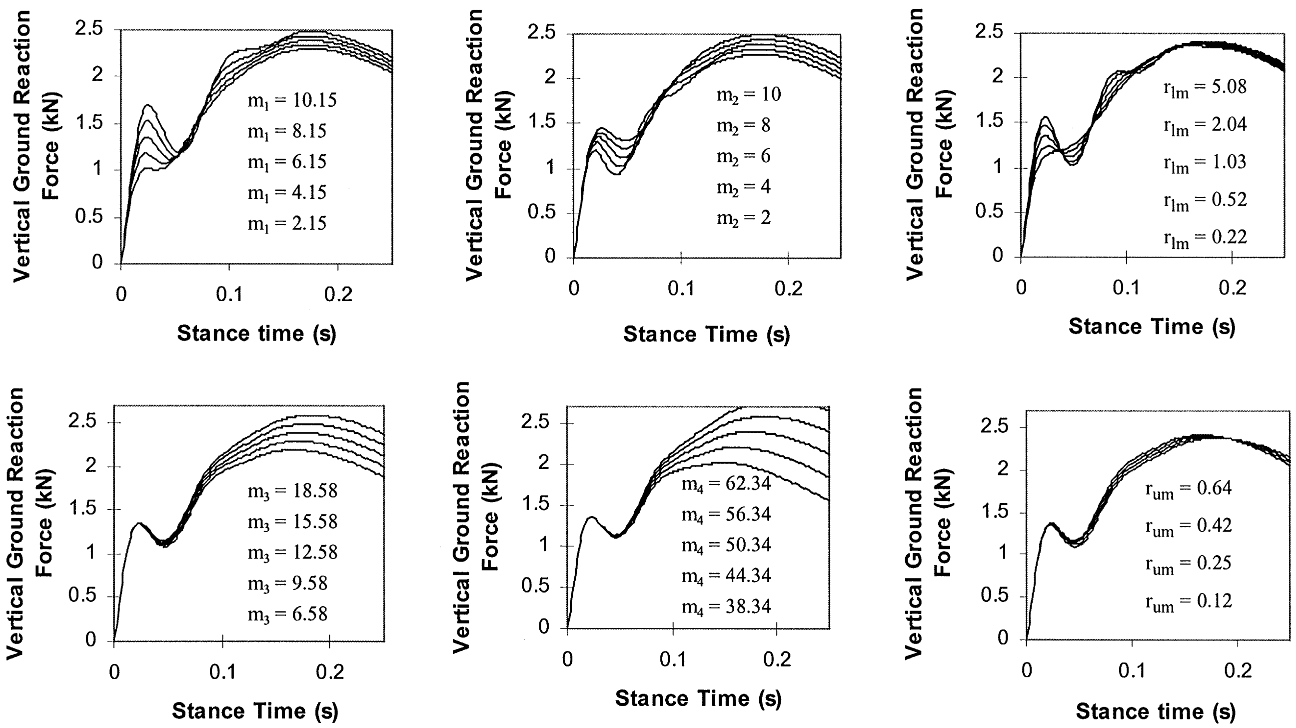

Fig. 3. Simulated ground reaction force–time history curves that show (a) increases in the lower ridig mass (m_1), (b) in the lower wobbling mass (m_2), and (c) in the mass ratio (r_{1m}) of lower body producing incremental increases in the impact force peak. The values of masses (kg) and mass ratio are listed in the figure form bottom up as related impact force peaks increase. Simulated ground reaction force–time history curves that show (d) increases int he upper rigid mass (m_3) (e) in the upper wobbling mass (m_4), and (f) in the mass ratio (r_{um}) of upper body producing minimum or no increase in the impact force peak. Great increases in the active force peak were produced by the increases in the m_3 and m_4. The values of masses (kg) and mass ratio were listed in the figure from bottom up as related active force peaks increase.

added weight caused increase of impact forces and a protective neuro-muscular response where impact forces did not change. The simulations in the current study with varied body masses were conducted without altering the spring and damper parameters of the system, therefore, represent a portion of the subjects with Newtonian response to the increased mass in the lower body (Bates et al., 1986). The results of the current study, on the other hand, indicated that body mass and mass distribution might strongly affect the control mechanism of the impact force for people who showed neuro-muscular response. It is suggested that body masses and mass distribution should be included in future studies on dynamics and neuro-muscular control mechanism of human running, especially when dealing with individual variations.

The results of the current simulations showed a minimum effect of variations in the upper body mass to the impact force peak. This phenomenon has been proposed in the past as Denoth (1986) defined an equivalent mass of running impact to be the ratio of the impact force peak versus the acceleration of the tibia. The further findings of the present study included that the upper wobbling mass had no effect on the impact force peak, and that the change of mass ratio in the upper body had small effect on the impact force peak due to a changed upper rigid mass. However, increases of the upper body masses showed strong influence on the active force peak during running. This increased active force peak need to be further examined since it may cause overuse injuries during running for people with over-weighted upper body.

The effect of body masses on the impact force peak in running was examined using a simplified model. The model consisted of upper and lower bodies with rigid and wobbling masses connected to each other. The results of simulations without altering the spring and damper parameters suggested an important effect of lower body masses on the impact force peak without neuro-muscular compensation. Future studies on the dynamics and neuro-muscular control of human running are required to take into consideration of the influence of individual variation in lower body mass and mass distribution.

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