Citation for published version:
Cazzola, D, Alberti, G, Ongaro, L & Minetti, AE 2014, 'The vertical excursion of the body visceral mass during vertical jumps is affected by specific respiratory maneuver', Human Movement Science, vol. 33, no. 1, pp. 369-380. https://doi.org/10.1016/j.humov.2013.10.008

DOI:
10.1016/j.humov.2013.10.008

Publication date:
2014

Document Version
Peer reviewed version

THE VERTICAL EXCURSION OF THE BODY VISCERAL MASS DURING VERTICAL JUMPS IS AFFECTED BY SPECIFIC RESPIRATORY MANOEUVRE

Dario Cazzola[1,4], Giampietro Alberti[2], Lucio Ongaro[3], Alberto E. Minetti[1]

[1]Laboratory of Physiomechanics, Department of Physiopathlogy and Transplantation, Faculty of Medicine, University of Milan, Italy, [2]Department of Sport, Nutrition and Health Sciences, Faculty of Exercise and Sports Sciences, University of Milan, [3]Faculty of Exercise and Sports Sciences, University of Milan, [4]Sport, Health and Exercise Science, Department of Health, University of Bath, UK.

Corresponding author: Dr Dario Cazzola, Research Officer - Sport, Health & Exercise Science - Department for Health | University of Bath - Applied Biomechanics Suite, 1.308. BA2 7AY, BATH (UK). Mobile (UK) ++44 (0) 7450820004. E-mail (University of Bath): dc547@bath.ac.uk. E-mail (home): dario.cazzola@me.com

ABSTRACT

Most of the modelling of body dynamics in sports assumes that every segment is ‘rigid’ and moves 'as a whole', although we know that uncontrolled wobbling masses exist and their motion should be minimized, both in engineering and biology. The visceral mass movement within the trunk segment potentially interferes with respiration and motion acts as locomotion or jumping. The aim of this paper is to refine and expand a previously published methodology to estimate that relative motion by testing its ability to detect the reduced vertical viscera excursion within the trunk. In fact, a respiratory-assisted jumping strategy is expected to limit viscera motion stiffening the abdominal content of the bouncing body. Six subjects were analysed, by using both inverse and direct dynamics, during repeated vertical jumps performed before and after a specific respiratory training period. The viscera excursion, which showed consistent intra-individual time courses, decreased by about 30% when the subjects had familiarized with the trunk-stiffening manoeuvre. We conclude that: 1) the present methodology proved to detect subtle visceral mass movement within the trunk during repetitive motor acts and, particularly, 2) a newly proposed respiratory manoeuvre/training devoted to stiff the trunk segment can reduce its vertical displacement.

1. INTRODUCTION

In biomechanical studies of human and animal motion and locomotion, the body is often simplified as composed by a number of rigid segments. From the location of those segments in 3D space, many important variables such as the body centre of mass (BCoM), the related internal and external mechanical work (Willems, Cavagna, & Heglund, 1995) are calculated to infer the characteristic dynamics of movement (A. E. Minetti, Cisotti, & Mian, 2011; Saibene & Minetti, 2003). Also, rotational parameters as joint net moments and segments inertial characteristics are based on the same "rigid body model". Unfortunately, such assumption can lead to experimental inaccuracies (Gao & Zheng, 2008; Leardini, Chiari, Della Croce, & Cappozzo, 2005). For this reason specific wobbling mass models have been proposed (Gruber, Ruder, Denoth, & Schneider, 1998; Yue & Mester, 2002) to improve and to refine experimental results especially during impacts (Gunther, Sholukha, Kessler, Wank, & Blickhan, 2003; M. T. G. Pain & Challis, 2004), in the attempt to enhance the description of the complex mechanical behaviour of the human body by including the contribution of soft parts. This approach allows quantification of the soft tissue deformation and displacement as a consequence of the impact forces transmission along the body (Challis & Pain, 2008; Wakeling & Nigg, 2001) during walking (Chen, Mukul, & Chou, 2011), running (Boyer & Nigg, 2007) and jumping (Gittoes, Brewin, & Kerwin, 2006; Mills, Scurr, & Wood, 2011). Soft tissue and viscera motion can also affect the external work of level and gradient walking (DeVita, Helseth, & Hortobagyi, 2007; Zelik & Kuo, 2010) and of running economy and stability (Daley & Usherwood, 2010). It as even be proposed that a suitable muscle-tuned control of that collateral effect could minimize the overall energy dissipation (Friesenbichler, Stirling, Federolf, & Nigg, 2011).

Thus, soft tissue and viscera movement has to be considered as a non-negligible factor in modelling optimization strategies and in experimental methodology, also in relation to the potential mechanical interaction with the rest of the body. For example, several authors have just pointed out the role of the visceral mass movement (within the trunk) in the locomotor-respiratory coupling during trotting and galloping in quadrupeds (Alexander, 1993; Bramble & Carrier, 1983; Simons, 1999). A similar condition occurs in humans, where some locomotor-respiratory coupling in running (McDermott, Van Emmerik, & Hamill, 2003) and walking (Rassler & Kohl, 1996) reflects the influence on the diaphragm function of the transient axial acceleration of abdominal viscera (Brown, Lee, & Loring, 2004; Loring, Lee, & Butler, 2001; Wilson & Liu, 1994). A very simple experiment illustrates this point: whoever tries to breath out-of-phase with respect to the spontaneous pattern during repeatedly jumping in place feels a great discomfort in achieving such a goal, mainly because respiratory muscles have to fight against the volume changes imposed by the jump-induced vertical accelerations of the visceral piston within its container.

In addition to the coupling between a cyclic activity as locomotion and respiration, there are other movements where the visceral mass displacement can play a role. In sport activities as volleyball, basketball or athletics, where jumping efficacy or horizontal-to-vertical velocity conversion are crucial (Yu & Hay, 1996), it is conceivable that controlling the wobbling mass could potentially avoid discomfort and energy dissipation associated to adverse oscillations, by also lowering workload perception (Bonsignore, Morici, Abate, Romano, & Bonsignore, 1998) or enhancing the jump performance. In this respect training techniques have been suggested to reduce the amplitude of that movement (Caufriez, 2005; Kapandji, 1977; Lumb,

2005) or even to obtain a beneficial influence on BCoM trajectory during the motion cycle.

A few years ago, a methodology using both 3D motion capture and platform dynamometry was proposed to infer the movement of the visceral mass during cyclic motor acts (A. Minetti & Belli, 1994). In short, by comparing the movement of the container (i.e. the rigid, multi-segment body) assessed by motion analysis, to the displacement of the 'true' BCoM, evaluated by double integration of the net vertical ground reaction force, it was possible to quantify the relative motion of the visceral mass within the trunk.

The aim of this paper was to apply that method to test whether a novel jumping technique, based on stiffening both chest and abdominal walls by means of a particular respiratory manoeuvre, was associated to the expected reduction in the visceral mass vertical displacement within the trunk. That would represent the first experimental evidence that the effects of a voluntary pattern of respiratory muscles activation during jumping can be accurately measured with a non-invasive approach.

2. Materials and Methods

2.1 Experimental Protocol

Six subjects (age 23.3 ± 2.5, trunk length 0.570 ± 0.110 m, weight 659.4 ± 53.0 N) were selected to jump in two different sessions on a force platform (model 9281C, Kistler, CH) measuring the vertical GRF synchronized with a six-camera motion capture system (Vicon MX, Oxford Metrics, UK). All the subjects were students from the Sport Science Faculty (University of Milan), chosen for their motor/jumping skill. The institutional ethics committee had approved all the methods and procedures, and subjects gave their informed consent prior to the experiments.

The platform signal was sampled at 1200 Hz, while the optoelectronic system captured frames at 400 Hz. The human body was modelled as a series of 14 linked, rigid body segments: 18 reflective markers (radius = 14 mm) were placed bilaterally on anatomical landmarks (Figure 1), nine on each side of the body (Mian, Thom, Ardigò, Narici, & Minetti, 2006), while 4 'technical-markers' were placed on the estimated centre of mass position of pectoral muscles, and right and left abdomen surface. Segment mass fraction and proximal distance of the centre of mass were taken from Dempster (Dempster, Gabel, & Felts, 1959).

The experiment consisted of two sessions, which were made up of 5 trials containing 15 consecutive jumps each, and spaced out by an adequate recovery period between trials. During the first session, the subjects jumped barefoot, with the hand on their hips, without any advice, to facilitate a natural jump execution. The second experimental session took place according to the same protocol after a training period of one month in which the subjects followed a specific learning progression devoted to jump in the “controlled” way (see below). Before the second session, the specific respiration technique and muscle contraction skills were tested on every subject: airflow was measured with a heated Fleisch pneumotachograph (HS Electronics, March-Hugstetten, Germany) connected to a facial mask and a differential pressure transducer (Validyne MP45, Northridge, CA). The activity of rectus and obliquus abdominis muscles was recorded via surface EMG (model ICP511, Grass Technologies, US), and the rectified EMG signal was filtered by 2^{th} order low-pass Butterworth filter with cut-off frequency of 6 Hz (Clancy, Morin, & Merletti, 2002). Both the signals were sampled at 1200 Hz by a 16-bit analog to-digital converter, and stored on a desk computer. Volume changes (V) were obtained by numerical integration of the digitized airflow signal, after calibration of the measuring apparatus by means of a graded cylinder and a metronome.

2.2 'Controlled' Jumping Technique

The training technique suggested in this study was designed according to the idea that by predominantly using 'low' diaphragmatic respiration, the visceral mass could be increasingly compacted towards the pelvis (Calais-Germain, 2005). With the spine in the physiological upright posture, a proper contraction activity of the abdominal wall/pelvic floor muscles avoids the forward displacement of the compressed viscera, improves the stiffness of the abdominal belt and, consequently, of the whole body structure (Le Boulch, 1973). This is achievable through a limited pelvis anteversion position, the preparatory low diaphragmatic inspiration (Figure 2a), and the simultaneous dorsum-lumbar filling caused by an intra-abdominal pressure increase, which is amplified by the forced expiration during the impact phases (Caufriez, 2005; Kapandji, 1977). Further details about the jumping/breathing technique and training can be obtained from co-authors LO and GA. In Figure 2b the EMG activity of rectus and obliquus abdominis muscles, together with the expired volume, are shown during normal and 'controlled' jumps.

2.3 Mechanical model

The method presented by Minetti and Belli is based on a model made up of a container with mass M, incorporating a hidden mass m (the visceral content), which oscillates periodically in the vertical or horizontal direction. In line with the original paper, we considered just vertical motion but included an 'external' wobbling mass

(m_e), representing mainly pectoral muscles and abdominal wall, as part of the container (see Figure 3). The new equation of motion is:

$$(M + m + m_e)\ddot{y}_{CoM}(t) = F_v(t) - (M + m + m_e)g \qquad (1)$$

which results from the system of equations:

$$\begin{cases} M\ddot{y}_1(t) = F_v(t) - Mg - f_v(t) - f_e(t) \\ m\ddot{y}_2(t) = f_v(t) - mg \\ m_e\ddot{y}_3(t) = f_e(t) - m_e g \end{cases} \qquad (2)$$

where F_v is the vertical component of GRF, f_v and f_e are vertical forces (unknown) exerted by the internal and 'external' masses, and y_1, y_2 and y_3 are distances from ground level of the container, visceral mass and external mass.

In literature, the magnitude of the internal visceral mass ‘m’ is estimated to be 16% of body mass (Martin, Janssens, Caboor, Clarys, & Marfell-Jones, 2003), while the external wobbling mass ‘m_e’ is evaluated to be 4% of body mass (Burkhart, Arthurs, & Andrews, 2008).

2.4 Data processing

A bespoke written software (LABVIEW 8.6, National Instrument, US) was developed to calculate the visceral mass vertical displacement, as shown in the equation (3),

$$s(t) - s_0 = \frac{(M+m+m_e)}{m}\left\{\left[\int_0^t\left(\int_0^t\left(\frac{F_v(t)}{M+m+m_e} - g\right)dt\right)dt - \frac{t}{T}\int_0^T\left(\int_0^t\left(\frac{F_v(t)}{M+m+m_e} - g\right)dt\right)dt\right] - \left(\frac{M+m}{M+m+m_e}\right)[y_1(t) - y_1(0)] - \left(\frac{m_e}{M+m+m_e}\right)[y_3(t) - y_3(0)]\right\}$$

(3)

where “T” is the movement period and “t” the progressive time.

This method and its algorithm were validated by loading in our program the kinetic data obtained from a simulation software (Visual Nastran 4D, MSC Software) of a known mechanical model (oscillating cylinder containing a sphere linked to the ceiling by a spring).

The developed software automatically recognized and isolate every jump (jump cycle = time between two subsequent BCoM peaks), double integrated (trapezoidal rule) the net GRF, and downsampled displacement data from 1200 Hz to 400 Hz to match the sampling rate of the motion capture system. GRF signal was shifted backward to cover a time gap ($=2 \cdot \Delta t/2=\Delta t$) due to double integration, to synchronize these data with kinematic acquisition. Force signal and kinematic data were filtered forward and backward by a 3rd order zero-lag low-pass Butterworth filter with cut-off frequency of 30 Hz (Bisseling & Hof, 2006). The frequency of the input signal (GRF), f_{GRF}, was used to compare the dynamics of subjects’ jumps (Boyer & Nigg, 2007) and its value was estimated by using the input peak value of the F_v, and the average loading rate between the 20% and 80% of the impact phase ($G_{v,ave}$), as:

$$f_{GRF} = \frac{1}{2(F_v/G_{v,ave})}$$

3. RESULTS

The biomechanical model chosen in this work allows an accurate BCoM estimation in locomotion (Halvorsen, Eriksson, Gullstrand, Tinmark, & Nilsson, 2009), and its adoption in jumping shows an error comparable to the literature. Indeed, two validation indices were estimated during the flight phase of the jumps: AV_1 (m/s^2) index represents an estimation of the gravity constant acceleration (g), expected to be 9.81 m/s^2, while AV_2 (m) index is defined as the root mean square error among the model estimated and matched ballistic centre of mass trajectory (Rabuffetti & Baroni, 1999). Their overall mean values and s.d. are respectively AV_1 (m/s^2) = -9.836 ± 0.027, AV_2 (m) = 0.003 ± 0.002.

In Table 1 the results of all the experiments are shown. The visceral mass (VMD), pectoral and abdomen external mass displacements (EMD) are represented as relative to the BCoM. The VMD, for all the subjects, measured during normal jumps (0.069 ± 0.020 m), is significantly higher ($p < 0.05$, paired t-test), than in controlled jumps (0.053 ± 0.018 m). The average time courses of normal and controlled VMD are shown in Figure 4, while the mean individual curves of partecipants are displayed in Figure 5.

For all the subjects, VMD shows a different pattern with respect to the container displacement both in normal and in controlled jumps (Figure 4), with a detectable phase shift between the curves. A paired t-test shows no significant difference of time shift, both during the aerial (normal 50.6 ± 10.4 ms – controlled 49.3 ± 9.4 ms) and landing (normal 51.2 ± 14.4 ms - controlled 49.8 ± 8.8 ms) phases, confirming a constant phase shift in both jumping techniques. A local maximum in visceral mass displacement ($\dot{s}(t) = 0$) is detectable at about 40-45% of jump period (time between two subsequent BCoM peaks) (Figure 4) and could be classified as a typical artefact of the foot impact on the force platform (Bisseling & Hof, 2006). The pectoral and abdominal EMD values show no significant difference in the two jumping techniques (paired t-test), but the pectoral EMD is significantly larger ($p<0.05$, paired t-test) than the abdomen EMD in both techniques (Figure 6).

Pectoral and abdomen EMD show a different pattern with respect to BCoM oscillation and VMD. Finally, a non-significant difference of f_{GRF}, jumping frequency (f_{jump}), BCoM vertical excursion and contact time (t_c) between the techniques (Table 1), for all the subjects, reveals a comparable dynamic and kinematic of normal and controlled jumps.

4. Discussion

The aim of this investigation was to test the effect of a combined respiratory/jumping strategy, properly designed for compacting viscera in the abdominal cavity, in limiting the vertical viscera motion during vertical jumps. Applying a previously developed method (A. Minetti & Belli, 1994), by concurrently using inverse and direct dynamics, we revealed that such a strategy reduced the vertical excursion up to 30%, with potential increases of the overall stiffness of the human trunk/body.

The VMD mean value measured was comparable with the literature: few quantitative analyses were conducted mostly anatomically (Beillas, Lafon, & Smith, 2009) or in slow-dynamic condition (Hostettler, Nicolau, Remond, Marescaux, & Soler, 2010), where vertical viscera motion was found to range between 0.03 m and 0.07 m. Only Minetti & Belli reported a value related to submaximal repeated jumps (0.08 m), while Boussuges and collaborators (Boussuges, Gole, & Blanc, 2009) set the limit of vertical displacement on maximal diaphragm motion (0.070 ± 0.011 m).

Regarding to the ‘controlled’ technique execution, experimental evidences of higher abdominal muscle activation and comparable expiration volume (Figure 2) proved that a voluntary diaphragm activation can be inferred: the volume of expired air during the controlled jump sequence was small and comparable with the normal jump, despite of a higher activation of expiratory muscles (obliquus and rectus abdominis), implying that the diaphragm applied an opposite force to contrast the rising viscera. In terms of interaction between respiration and movement, our results show that muscles not directly involved in jumping could affect body dynamics, and stress their potential effect on motor acts where locomotor/respiratory coupling-ratios can occur.

In the literature several authors have already speculated about frequency and phase coupling between respiratory and locomotory rhythms as affected by training (Bernasconi & Kohl, 1993) or workload (Rassler & Kohl, 1996), but no one provided evidences of voluntary control of internal body dynamics through specific respiration techniques, synchronously performed with body CoM oscillations. Only McDermott (McDermott, et al., 2003), by investigating the relationship between locomotor/respiratory coupling and training level, found that expert runners were particularly skilled in synching their coupling during speed changes. Therefore, from the energetic point of view, these interactions should be controlled to avoid energy losses resulting in some extra-mechanical work done by muscles, and the time delay calculated between BCoM and VMD curves in this investigation, reinforces this hypothesis. In fact, the ‘economy’ of bouncing locomotion, such as running or skipping, could be influenced and the mechanical external work calculated from kinematically measured CoM displacement could be refined by adding viscera contribution (Daley & Usherwood, 2010). While this is supposed to be a small adjustment in normal subjects, any deviation from a mesomorphic body such as obese patients with relevant internal and external wobbling masses would involve a more substantial correction of the inverse dynamics approach. In this way the proposed respiratory strategy could give potential benefits in terms of movement performance and the non-invasive method described could be easily adopted.

In terms of data processing the previous method (Minetti & Belli, 1994) has been refined: kinematic sampling frequency has been quadrupled (400 Hz) and chosen as a submultiple of the dynamometric signal to facilitate synchronization, the signals were accurately aligned (double integration time gap), and the mathematical model was validated with physics laboratory simulation software. Besides, the method still suffered of inaccuracies due to: 1) the rigid body model assumption (Cappozzo, Della Croce, Leardini, & Chiari, 2005; Chiari, Della Croce, Leardini, & Cappozzo, 2005) originating troublesome theoretical interpretations of the results: the discrepancy between the BCoM estimates from direct and inverse dynamics is considered as an indirect evidence of viscera motion, but this could be partially the results of experimental inaccuracies, 2) the “skin marker artefact” (Cappozzo, Catani, Leardini, Benedetti, & Croce, 1996), which particularly affects movements with considerable joint rotation as sit-to-stand (Kuo, et al., 2011) or locomotion (Akbarshahi, et al., 2010) rather than vertical jumps with the arms blocked on the trunk, 3) the “soft tissue motion artefact” (Gruber, et al., 1998; Leardini, et al., 2005), which can be assessed by accelerometers (Kitazaki & Griffin, 1995) or by adding extra markers for the oscillating body parts, at the cost of a more complex biomechanical model. The 4 'technical' markers introduced here, positioned on the estimated centre of mass of the most visible and bulky 'external' wobbling masses (pectorals and abdominal muscles), allowed their movement to contribute to refine VMD estimation. This simplified approach does not completely compensate for the rigid body assumption inaccuracies and cannot separate viscera from limbs soft tissues contribution (Gunther, et al., 2003), but it constitutes an acceptable trade-off between ideal VMD estimation and practical feasibility.

A further variable affecting VMD and EMD measure is the muscle tuning during jumping: the ‘controlled jump’ is comparable with a tuned landing thanks to an higher pectoral and abdominal muscles activation and could decrease the absolute and relative acceleration of the soft tissue compartments (Boyer & Nigg, 2006). Even though a further frequency analysis of external masses acceleration signal (not measured in this work) could reveal soft tissues vibrational changes between the techniques, pectoral and abdominal EMD are not significantly different (Table 1), and their patterns are similar in normal and controlled jumps (Figure 6). This is probably due to similar pectoral-muscle activation in both techniques, and to a peculiar muscle tuning effect on abdominal soft tissue: actually its vibration could be less influenced by muscle contraction than other soft tissues (upper/lower limbs) because of its anatomical characteristics and local physical constrains.

To date, soft tissues influences has already been investigated in locomotion (DeVita, et al., 2007; Zelik & Kuo, 2010) and in jump landing (Gittoes, et al., 2006; M. T. Pain & Challis, 2006), though its role still needs to be ultimately assessed. In this work, even if there are several limitations, we compared two refined estimations of the most influent soft tissue (viscera) motion in a simple motor task, repeatedly executed in the same experimental condition. Indeed, subjects executed comparable jumps considering the jumping frequency (fjump), contact time (tc), frequency of input force (fGRF) and the performance (body CoM vertical excursion). These evidences help to minimize systematic and random errors, showing a de-noised measure of viscera vertical excursion.

In conclusion, the combination of the inverse/direct dynamics method to measure viscera motion and a novel respiration assisted jumping technique reveals, for the first time, that the vertical displacement of the abdominal wobbling mass can be modulated also in dynamic condition. Moreover, it has been demonstrated that the accuracy of this refined method is adequate to detect, with a non-invasive approach, the effects of internal forces on the kinematic of the visceral mass and could be adopted to evaluate those their impact in sport biomechanics and locomotion energetics. The results and the proposed jumping strategy could then constitute a pre-requisite for further studies assessing the potential performance enhancement in a variety of motor acts.

ACKNOWLEDGEMENTS

The authors thank Dr Matteo Pecchiari for his invaluable experience in respiration physiology, and Dr Ezio Preatoni for his precious suggestions during paper revision.

REFERENCES

Akbarshahi, M., Schache, A. G., Fernandez, J. W., Baker, R., Banks, S., & Pandy, M. G. (2010). Non-invasive assessment of soft-tissue artifact and its effect on knee joint kinematics during functional activity. *J Biomech, 43*, 1292-1301.

Alexander, R. M. (1993). Breathing while trotting. *Science, 262*, 196-197.

Beillas, P., Lafon, Y., & Smith, F. W. (2009). The effects of posture and subject-to-subject variations on the position, shape and volume of abdominal and thoracic organs. *Stapp Car Crash J, 53*, 127-154.

Bernasconi, P., & Kohl, J. (1993). Analysis of co-ordination between breathing and exercise rhythms in man. *J Physiol, 471*, 693-706.

Bisseling, R. W., & Hof, A. L. (2006). Handling of impact forces in inverse dynamics. In *J Biomech* (Vol. 39, pp. 2438-2444). United States.

Bonsignore, M. R., Morici, G., Abate, P., Romano, S., & Bonsignore, G. (1998). Ventilation and entrainment of breathing during cycling and running in triathletes. *Med Sci Sports Exerc, 30*, 239-245.

Boussuges, A., Gole, Y., & Blanc, P. (2009). Diaphragmatic motion studied by m-mode ultrasonography: methods, reproducibility, and normal values. In *Chest* (Vol. 135, pp. 391-400). United States.

Boyer, K. A., & Nigg, B. M. (2006). Muscle tuning during running: implications of an un-tuned landing. *J Biomech Eng, 128*, 815-822.

Boyer, K. A., & Nigg, B. M. (2007). Quantification of the input signal for soft tissue vibration during running. In *J Biomech* (Vol. 40, pp. 1877-1880). United States.

Bramble, D. M., & Carrier, D. R. (1983). Running and breathing in mammals. *Science, 219*, 251-256.

Brown, R. E., Lee, H. T., & Loring, S. H. (2004). Airflow synchronous with oscillatory acceleration reflects involuntary respiratory muscle activity. In *Respir Physiol Neurobiol* (Vol. 140, pp. 265-282). Netherlands: 2004 Elsevier B.V.

Burkhart, T. A., Arthurs, K. L., & Andrews, D. M. (2008). Reliability of upper and lower extremity anthropometric measurements and the effect on tissue mass predictions. In *J Biomech* (Vol. 41, pp. 1604-1610). United States.

Calais-Germain, B. (2005). *Respiration anatomie-geste respiratoire*.

Cappozzo, A., Catani, F., Leardini, A., Benedetti, M. G., & Croce, U. D. (1996). Position and orientation in space of bones during movement: experimental artefacts. *Clin Biomech (Bristol, Avon), 11*, 90-100.

Cappozzo, A., Della Croce, U., Leardini, A., & Chiari, L. (2005). Human movement analysis using stereophotogrammetry. Part 1: theoretical background. *Gait Posture, 21*, 186-196.

Caufriez, M. (2005). *Respiration anatomie-geste respiratoire* (Editions DesIris ed.).

Challis, J. H., & Pain, M. T. (2008). Soft tissue motion influences skeletal loads during impacts. In *Exerc Sport Sci Rev* (Vol. 36, pp. 71-75). United States.

Chen, S. J., Mukul, M., & Chou, L. S. (2011). Soft-tissue movement at the foot during the stance phase of walking. In *J Am Podiatr Med Assoc* (Vol. 101, pp. 25-34). United States.

Chiari, L., Della Croce, U., Leardini, A., & Cappozzo, A. (2005). Human movement analysis using stereophotogrammetry. Part 2: instrumental errors. *Gait Posture, 21*, 197-211.

Clancy, E. A., Morin, E. L., & Merletti, R. (2002). Sampling, noise-reduction and amplitude estimation issues in surface. *J Electromyogr Kinesiol, 12*, 1-16.

Daley, M. A., & Usherwood, J. R. (2010). Two explanations for the compliant running paradox: reduced work of bouncing viscera and increased stability in uneven terrain. In *Biol Lett* (Vol. 6, pp. 418-421). England.

Dempster, W. T., Gabel, W. C., & Felts, W. J. (1959). The anthropometry of the manual work space for the seated subject. *Am J Phys Anthropol, 17*, 289-317.

DeVita, P., Helseth, J., & Hortobagyi, T. (2007). Muscles do more positive than negative work in human locomotion. In *J Exp Biol* (Vol. 210, pp. 3361-3373). England.

Friesenbichler, B., Stirling, L. M., Federolf, P., & Nigg, B. M. (2011). Tissue vibration in prolonged running. In *J Biomech* (Vol. 44, pp. 116-120). United States: 2010 Elsevier Ltd.

Gao, B., & Zheng, N. N. (2008). Investigation of soft tissue movement during level walking: translations and rotations of skin markers. In *J Biomech* (Vol. 41, pp. 3189-3195). United States.

Gittoes, M. J., Brewin, M. A., & Kerwin, D. G. (2006). Soft tissue contributions to impact forces simulated using a four-segment wobbling mass model of forefoot-heel landings. In *Hum Mov Sci* (Vol. 25, pp. 775-787). Netherlands.

Gruber, K., Ruder, H., Denoth, J., & Schneider, K. (1998). A comparative study of impact dynamics: wobbling mass model versus rigid body models. In *J Biomech* (Vol. 31, pp. 439-444). United States.

Gunther, M., Sholukha, V. A., Kessler, D., Wank, V., & Blickhan, R. (2003). Dealing with skin motion and wobbling masses in inverse dynamics. *J Mech in Med and Bio (JMMB), 3*, 309-335.

Halvorsen, K., Eriksson, M., Gullstrand, L., Tinmark, F., & Nilsson, J. (2009). Minimal marker set for center of mass estimation in running. *Gait Posture, 30*, 552-555.

Hostettler, A., Nicolau, S. A., Remond, Y., Marescaux, J., & Soler, L. (2010). A real-time predictive simulation of abdominal viscera positions during quiet free breathing. In *Prog Biophys Mol Biol*: 2010 Elsevier Ltd.

Kapandji, I. A. (1977). *Fisiologia articolare - Tronco e Rachide* (Vol. 3). Rome.

Kitazaki, S., & Griffin, M. J. (1995). A data correction method for surface measurement of vibration on the human body. *J Biomech, 28*, 885-890.

Kuo, M. Y., Tsai, T. Y., Lin, C. C., Lu, T. W., Hsu, H. C., & Shen, W. C. (2011). Influence of soft tissue artifacts on the calculated kinematics and kinetics of total knee replacements during sit-to-stand. *Gait Posture, 33*, 379-384.

Le Boulch, J. (1973). *L'éducation par le mouvement* (12éme èdition ed.). Paris.

Leardini, A., Chiari, L., Della Croce, U., & Cappozzo, A. (2005). Human movement analysis using stereophotogrammetry. Part 3. Soft tissue artifact assessment and compensation. *Gait Posture, 21*, 212-225.

Loring, S. H., Lee, H. T., & Butler, J. P. (2001). Respiratory effects of transient axial acceleration. *J Appl Physiol, 90*, 2141-2150.

Lumb, A. B. (2005). Nunn's Applied Respiratory Physiology. In (Sixth Editions ed., pp. 76-80).

Martin, A. D., Janssens, V., Caboor, D., Clarys, J. P., & Marfell-Jones, M. J. (2003). Relationships between visceral, trunk and whole-body adipose tissue weights by cadaver dissection. In *Ann Hum Biol* (Vol. 30, pp. 668-677). England.

McDermott, W. J., Van Emmerik, R. E., & Hamill, J. (2003). Running training and adaptive strategies of locomotor-respiratory coordination. *Eur J Appl Physiol, 89*, 435-444.

Mian, O., Thom, J., Ardigò, L., Narici, M., & Minetti, A. (2006). Metabolic cost, mechanical work, and efficiency during walking in young and older men. *Acta Physiol (Oxf), 186*, 127-139.

Mills, C., Scurr, J., & Wood, L. (2011). A protocol for monitoring soft tissue motion under compression garments during drop landings. In *J Biomech* (Vol. 44, pp. 1821-1823). United States: 2011 Elsevier Ltd.

Minetti, A., & Belli, G. (1994). A model for the estimation of visceral mass displacement in periodic movements. *J Biomech, 27*, 97-101.

Minetti, A. E., Cisotti, C., & Mian, O. S. (2011). The mathematical description of the body centre of mass 3D path in human and animal locomotion. In *J Biomech* (Vol. 44, pp. 1471-1477). United States: 2011 Elsevier Ltd.

Pain, M. T., & Challis, J. H. (2006). The influence of soft tissue movement on ground reaction forces, joint torques and joint reaction forces in drop landings. In *J Biomech* (Vol. 39, pp. 119-124). United States.

Pain, M. T. G., & Challis, J. H. (2004). Wobbling mass influence on impact ground reaction forces: A simulation model sensitivity analysis. *Journal of Applied Biomechanics, 20(3)*, 309-316.

Rassler, B., & Kohl, J. (1996). Analysis of coordination between breathing and walking rhythms in humans. In *Respir Physiol* (Vol. 106, pp. 317-327). Netherlands.

Saibene, F., & Minetti, A. E. (2003). Biomechanical and physiological aspects of legged locomotion in humans. *Eur J Appl Physiol, 88*, 297-316.

Simons, R. S. (1999). Running, breathing and visceral motion in the domestic rabbit (Oryctolagus cuniculus): testing visceral displacement hypotheses. *J Exp Biol, 202*, 563-577.

Wakeling, J. M., & Nigg, B. M. (2001). Soft-tissue vibrations in the quadriceps measured with skin mounted transducers. In *J Biomech* (Vol. 34, pp. 539-543). United States.

Willems, P. A., Cavagna, G. A., & Heglund, N. C. (1995). External, internal and total work in human locomotion. *J Exp Biol, 198*, 379-393.

Wilson, T. A., & Liu, S. (1994). Effect of acceleration on the chest wall. *J Appl Physiol, 76*, 1242-1246.

Yu, B., & Hay, J. G. (1996). Optimum phase ratio in the triple jump. In *J Biomech* (Vol. 29, pp. 1283-1289). United States.

Yue, Z., & Mester, J. (2002). A model analysis of internal loads, energetics, and effects of wobbling mass during the whole-body vibration. In *J Biomech* (Vol. 35, pp. 639-647). United States.

Zelik, K. E., & Kuo, A. D. (2010). Human walking isn't all hard work: evidence of soft tissue contributions to energy dissipation and return. In *J Exp Biol* (Vol. 213, pp. 4257-4264). England.

Figure 1: Human body modelled with 22 reflective markers and 14 segments: head (1), trunk (2), abdomen (4), right upper arm (5), left upper arms (6), right fore arm (7), left fore arm (8), right thigh (9), left thigh (10), right shank (11), left thigh (12), right foot (13), left foot (14), and pectoral muscles (3).

Figure 2: (a) Mechanism used to generate an intra-abdominal pressure that compacts the visceral mass: the subject after a combined deep diaphragmatic inspiration and contraction of the abdominal "press" increases the intra-abdominal pressure also executing progressive and short exhalations. The black arrows indicate: (1) The lowering of the diaphragm that pushes on the viscera during inspiration (downward-pointing white arrow); (2) The musculature of the abdominal "press", which contraction contributes to the elevation of intra-abdominal pressure (upward-pointing white arrows). (b) On the left the overall mean (normalized in respect of the maximal contraction value) and s.d of all the subjects, of rectus and obliquus abdominis muscle activation, in normal (light-grey) and controlled (dark-grey) jump are shown. The rectus and obliquus muscle activation is significantly higher in controlled jumps (* = $p < 0.01$). On the right the overall mean and s.d., of the expired volume (V) during a jump are shown. The expired volume is not significantly different between the techniques.

Figure 3: Model used for the estimation of visceral mass displacement: M is the container mass, m the internal visceral mass, and me is the external mass, while y1, y2 and y3 are distances from ground level and s=y2-y1. The whole system oscillates vertically and exerts a vertical ground reaction force Fv, while internal and external mass exerts a force f_v and f_e respectively on the container.

Figure 4: The overall mean curve of VMD (visceral mass displacement) in normal (grey solid line) and controlled (grey dashed line) jumps, and overall mean curve (controlled and normal) of body CoM (black solid line) are shown. All the curves are time-normalized with single jump duration (0-100%).

Figure 5: The mean of all the trials curves (5 trial of at least 15 jumps for every subject), presented with black bold line, and their variability (s.d. of all the trials curves), presented with light grey lines, are shown for both techniques (normal and controlled) for each subject (S1, S2, S3, S4, S5, S6). The curves are time-normalized with single jump duration.

Figure 6: The overall mean curve of pEMD (pectoral external mass displacement) in normal (black solid line) and controlled (black dashed line) jumps, the overall mean

curve of aEMD (abdominal external mass displacement) in normal (grey solid line) and controlled (grey dashed line) jumps, and the overall mean curve (controlled and normal) of body CoM (black dotted line). All the curves are time-normalized with single jump duration (0-100%). The pEMD and aEMD, for all the subjects, are not significantly different in the two techniques, but the pEMD is significantly higher ($p < 0.05$) than aEMD both in normal and in controlled jumps.

Table 1: The mean and s.d. values of (1) visceral mass displacement (VMD), (2) body CoM displacement (CoM), (3) pectorals (overall mean of right and left) external mas displacement (pEMD), (4) abdomen (overall mean of right and left) external mass displacement (EMD), (5) estimated input frequency (fGRF), (6) jumping frequency (f_{jump}) and (7) contact time (t_c) in “normal” and “controlled” jumps are presented for every subject.

Table 1

JUMP type	Subject	N		VMD (m)	CoM (m)	pEMD (m)	aEMD (m)	f_{GRF} (Hz)	f_{jump} (Hz)	t_c (s)
Normal	*S1*	76	Mean	0.073	0.209	0.030	0.016	7.13	2.40	0.106
			SD	0.015	0.019	0.008	0.007	0.69	0.02	0.003
	S2	70	Mean	0.089	0.347	0.042	0.026	6.79	1.66	0.114
			SD	0.005	0.049	0.009	0.008	0.51	0.07	0.003
	S3	85	Mean	0.059	0.168	0.031	0.010	8.42	1.96	0.101
			SD	0.005	0.007	0.010	0.006	0.30	0.13	0.003
	S4	85	Mean	0.056	0.216	0.029	0.018	7.67	2.09	0.109
			SD	0.008	0.026	0.008	0.008	0.53	0.06	0.006
	S5	71	Mean	0.102	0.311	0.040	0.024	7.21	1.82	0.098
			SD	0.005	0.013	0.008	0.009	0.33	0.03	0.001
	S6	90	Mean	0.051	0.137	0.049	0.041	6.76	2.65	0.067
			SD	0.006	0.008	0.010	0.010	0.67	0.10	0.002
	All	477	Mean	0.069	0.219	0.037	0.023	7.35	2.09	0.099
			SD	0.020	0.075	0.009	0.011	0.79	0.34	0.015
Controlled	*S1*	80	Mean	0.051	0.161	0.028	0.012	7.95	2.37	0.100
			SD	0.008	0.011	0.009	0.005	0.46	0.03	0.003
	S2	72	Mean	0.078	0.321	0.026	0.026	6.88	1.81	0.104
			SD	0.007	0.029	0.008	0.008	0.41	0.03	0.002
	S3	92	Mean	0.049	0.171	0.021	0.012	8.28	2.28	0.101
			SD	0.006	0.010	0.009	0.009	0.53	0.29	0.002
	S4	86	Mean	0.046	0.242	0.031	0.013	7.59	2.36	0.096
			SD	0.009	0.036	0.010	0.009	0.56	0.03	0.001
	S5	69	Mean	0.076	0.306	0.040	0.019	7.02	1.80	0.103
			SD	0.010	0.013	0.011	0.008	0.25	0.02	0.004
	S6	93	Mean	0.030	0.155	0.046	0.038	6.78	2.74	0.069
			SD	0.004	0.011	0.010	0.010	0.68	0.03	0.001
	All	492	Mean	0.053	0.217	0.032	0.020	7.46	2.21	0.097
			SD	0.018	0.069	0.009	0.010	0.77	0.35	0.012

Figure 1
Click here to download high resolution image

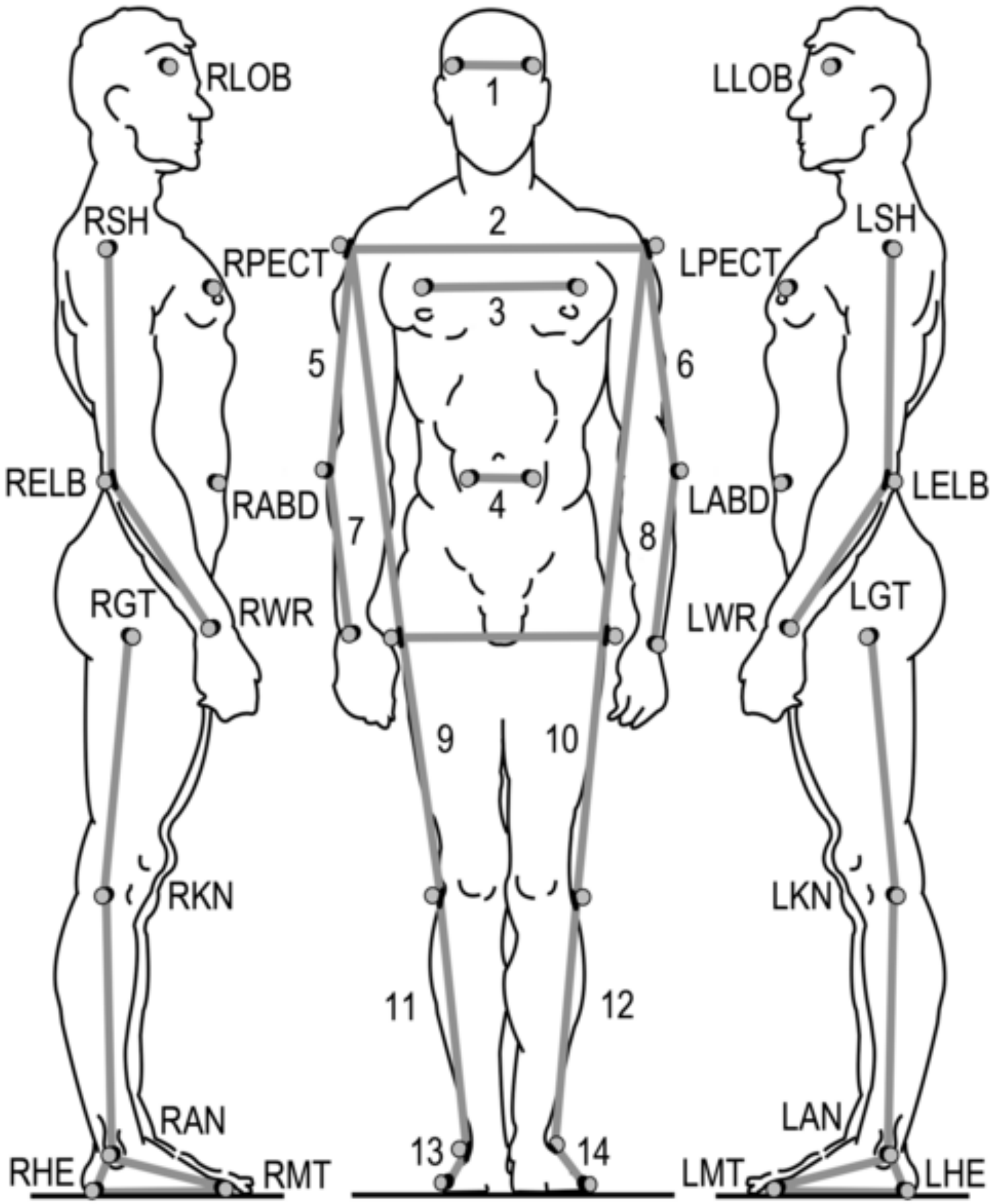

Figure 2
Click here to download high resolution image

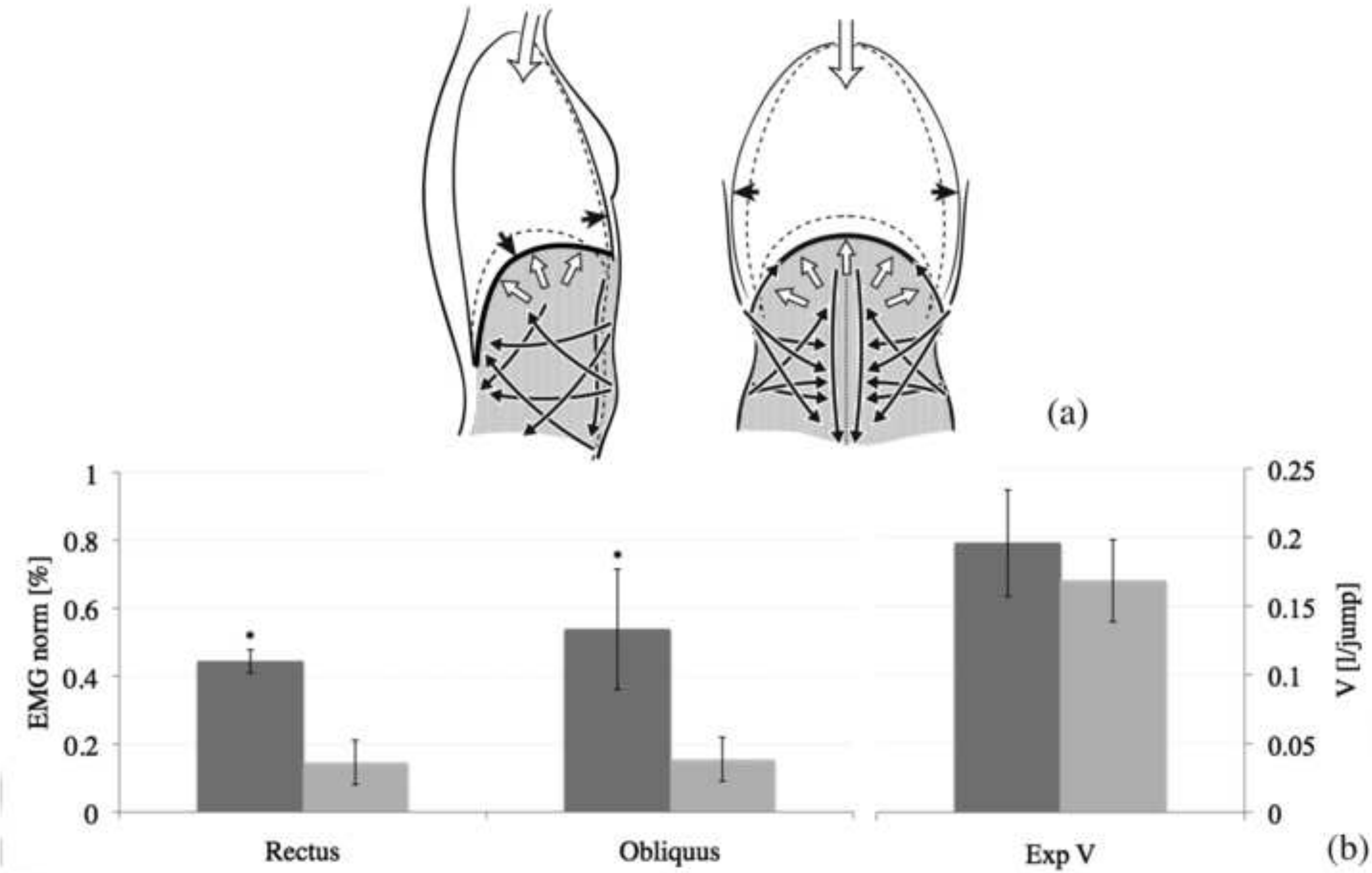

Figure 3
Click here to download high resolution image

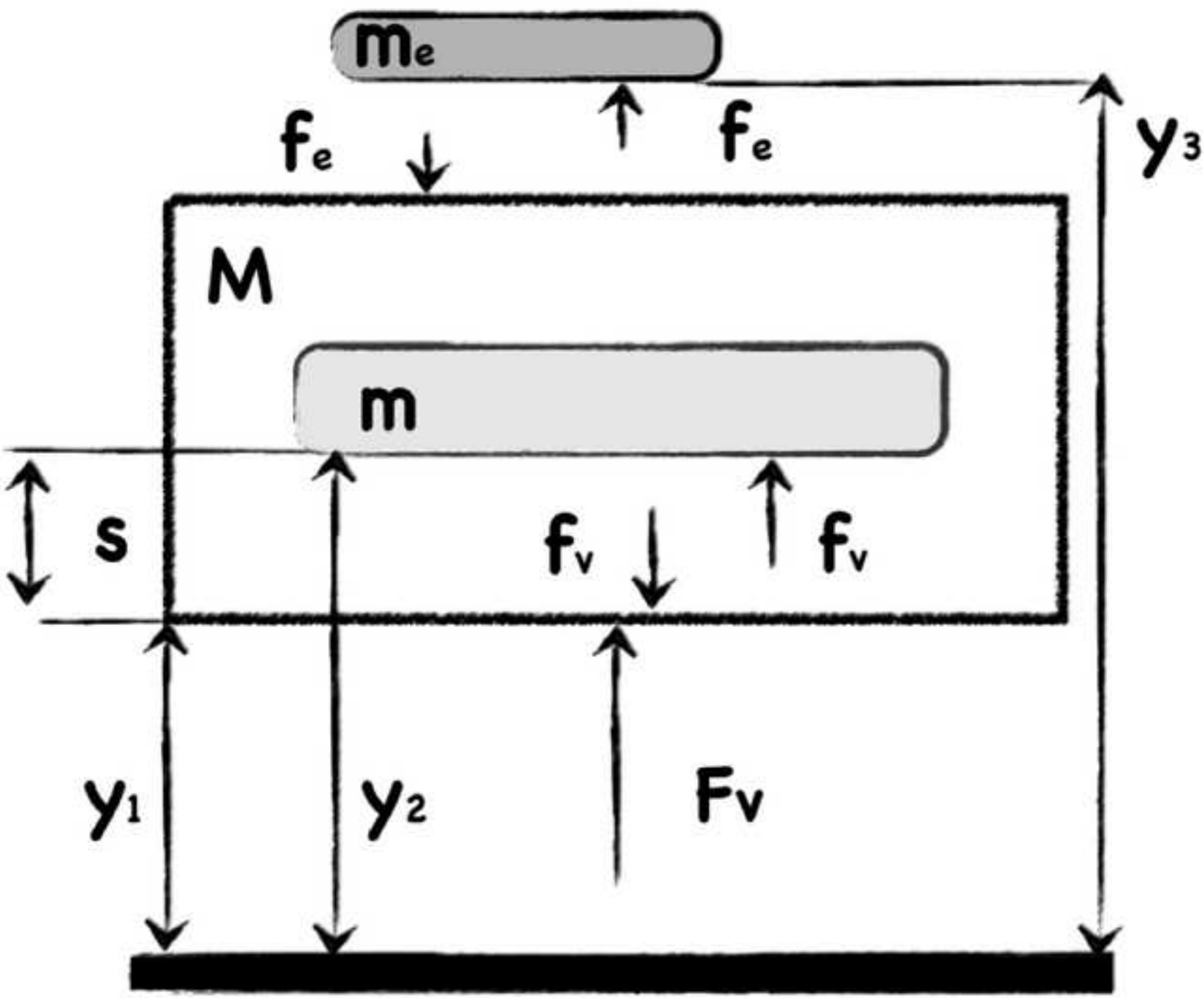

Figure 4

Click here to download high resolution image

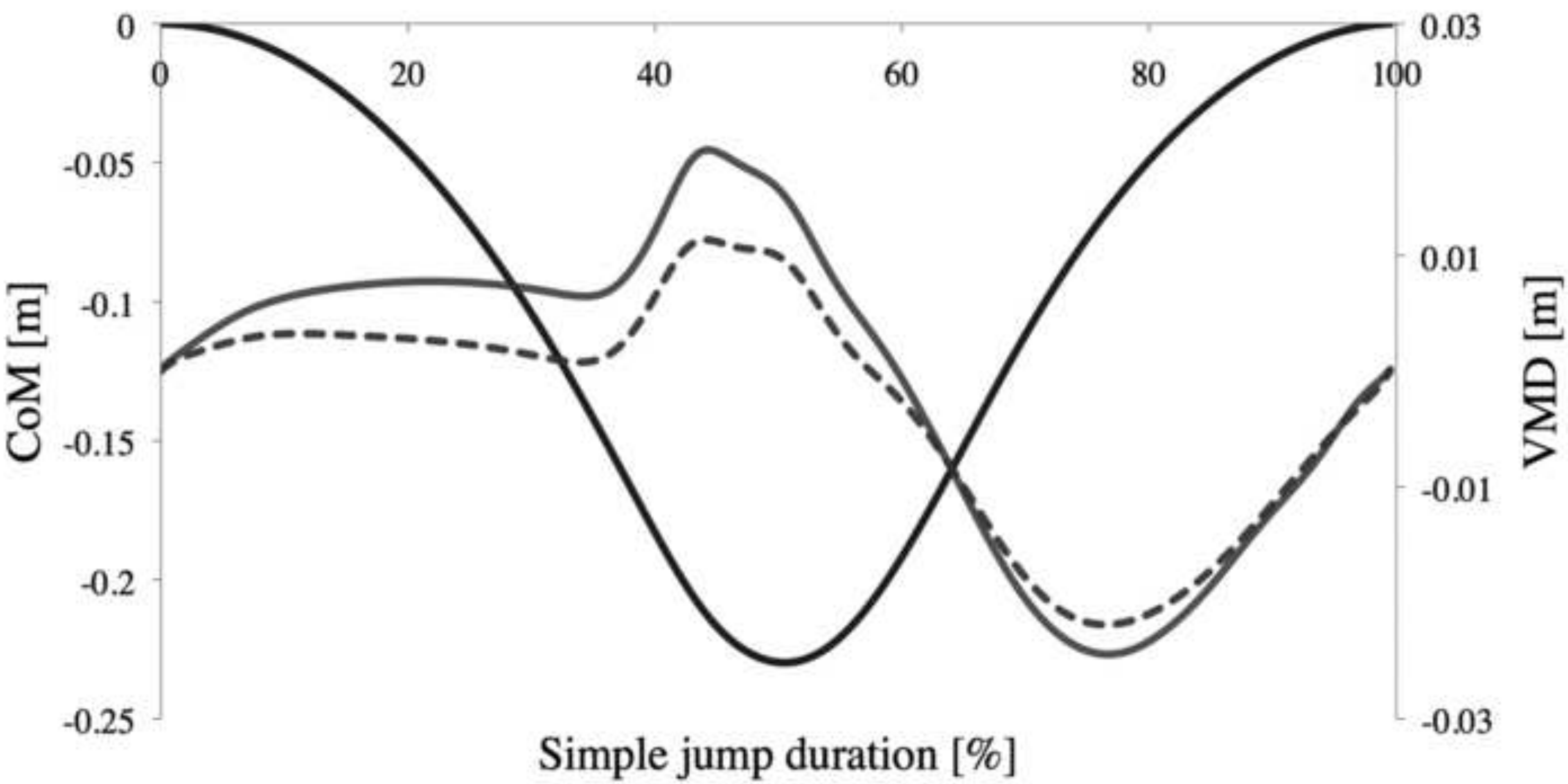

Figure 6

Click here to download high resolution image

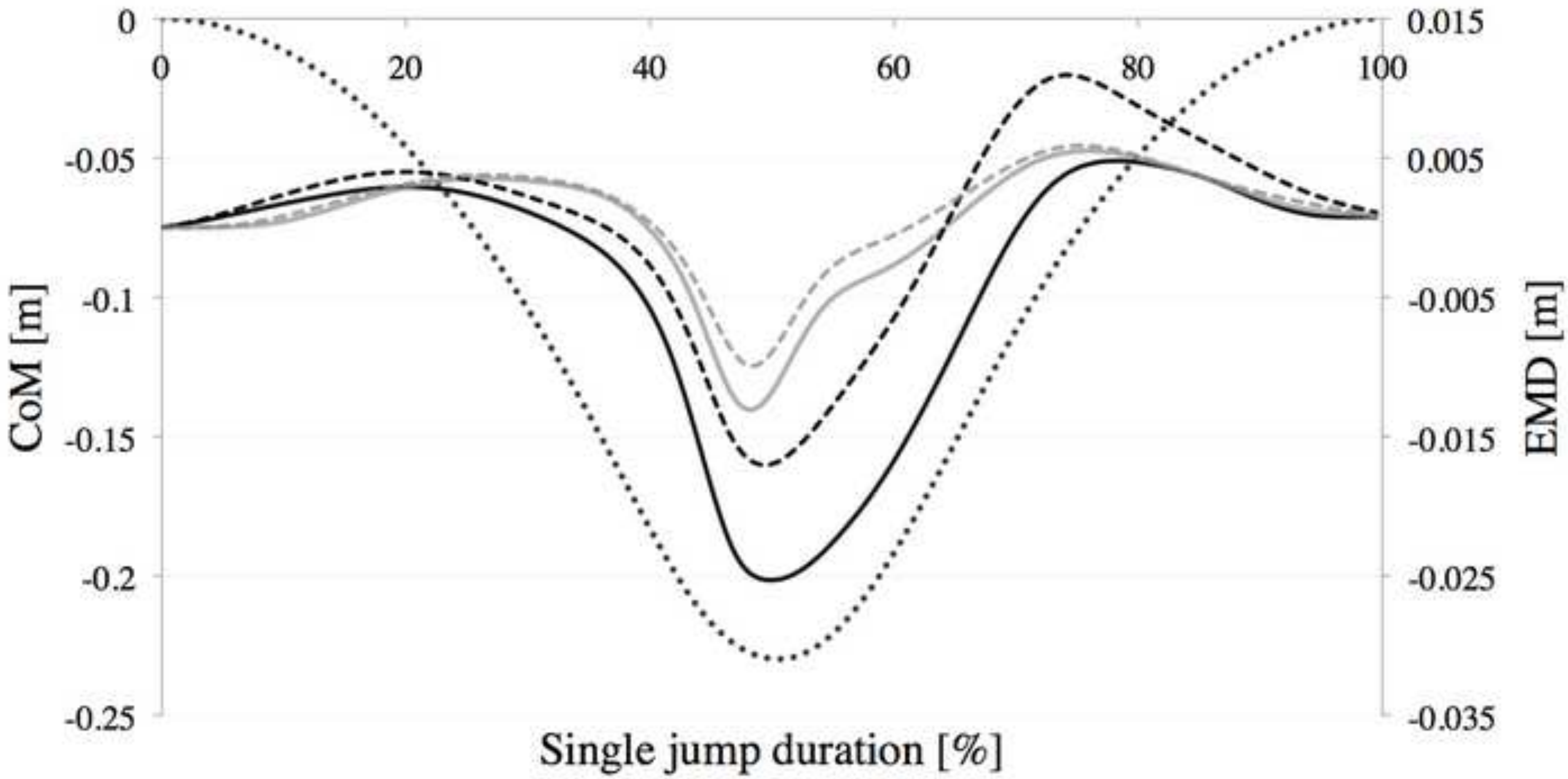

Figure 5
Click here to download high resolution image