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

4

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18

19 **ABSTRACT**

20 Most of the modelling of body dynamics in sports assumes that every segment is
21 'rigid' and moves 'as a whole', although we know that uncontrolled wobbling masses
22 exist and their motion should be minimized, both in engineering and biology. The
23 visceral mass movement within the trunk segment potentially interferes with
24 respiration and motion acts as locomotion or jumping. The aim of this paper is to
25 refine and expand a previously published methodology to estimate that relative
26 motion by testing its ability to detect the reduced vertical viscera excursion within the
27 trunk. In fact, a respiratory-assisted jumping strategy is expected to limit viscera
28 motion stiffening the abdominal content of the bouncing body. Six subjects were
29 analysed, by using both inverse and direct dynamics, during repeated vertical jumps
30 performed before and after a specific respiratory training period. The viscera
31 excursion, which showed consistent intra-individual time courses, decreased by about
32 30% when the subjects had familiarized with the trunk-stiffening manoeuvre. We
33 conclude that: 1) the present methodology proved to detect subtle visceral mass
34 movement within the trunk during repetitive motor acts and, particularly, 2) a newly
35 proposed respiratory manoeuvre/training devoted to stiff the trunk segment can
36 reduce its vertical displacement.

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41 1. INTRODUCTION

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

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

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

90 2005) or even to obtain a beneficial influence on BCoM trajectory during the motion
91 cycle.

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

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

105

106 2. MATERIALS AND METHODS

107 2.1 EXPERIMENTAL PROTOCOL

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

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

123 The experiment consisted of two sessions, which were made up of 5 trials containing
124 15 consecutive jumps each, and spaced out by an adequate recovery period between
125 trials. During the first session, the subjects jumped barefoot, with the hand on their
126 hips, without any advice, to facilitate a natural jump execution. The second
127 experimental session took place according to the same protocol after a training period
128 of one month in which the subjects followed a specific learning progression devoted
129 to jump in the “controlled” way (see below). Before the second session, the specific
130 respiration technique and muscle contraction skills were tested on every subject:
131 airflow was measured with a heated Fleisch pneumotachograph (HS Electronics,
132 March-Hugstetten, Germany) connected to a facial mask and a differential pressure
133 transducer (Validyne MP45, Northridge, CA). The activity of rectus and obliquus
134 abdominis muscles was recorded via surface EMG (model ICP511, Grass
135 Technologies, US), and the rectified EMG signal was filtered by 2th order low-pass
136 Butterworth filter with cut-off frequency of 6 Hz (Clancy, Morin, & Merletti, 2002).
137 Both the signals were sampled at 1200 Hz by a 16-bit analog to-digital converter, and
138 stored on a desk computer. Volume changes (V) were obtained by numerical

139 integration of the digitized airflow signal, after calibration of the measuring apparatus
140 by means of a graded cylinder and a metronome.

141

142 2.2 'CONTROLLED' JUMPING TECHNIQUE

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

157

158 2.3 MECHANICAL MODEL

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

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

$$165 \quad (M + m + m_e)\ddot{y}_{CoM}(t) = F_v(t) - (M + m + m_e)g \quad (1)$$

166 which results from the system of equations:

$$167 \quad \begin{cases} M\ddot{y}_1(t) = F_v(t) - Mg - f_v(t) - f_e(t) \\ m\ddot{y}_2(t) = f_v(t) - mg \\ m_e\ddot{y}_3(t) = f_e(t) - m_e g \end{cases} \quad (2)$$

168

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

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

176

177 2.4 DATA PROCESSING

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

180

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

181

182 (3)

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

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

188 The developed software automatically recognized and isolate every jump (jump cycle
189 = time between two subsequent BCoM peaks), double integrated (trapezoidal rule) the
190 net GRF, and downsampled displacement data from 1200 Hz to 400 Hz to match the
191 sampling rate of the motion capture system. GRF signal was shifted backward to
192 cover a time gap ($=2 \cdot \Delta t / 2 = \Delta t$) due to double integration, to synchronize these data
193 with kinematic acquisition. Force signal and kinematic data were filtered forward and
194 backward by a 3rd order zero-lag low-pass Butterworth filter with cut-off frequency of
195 30 Hz (Bisseling & Hof, 2006). The frequency of the input signal (GRF), f_{GRF} , was
196 used to compare the dynamics of subjects’ jumps (Boyer & Nigg, 2007) and its value
197 was estimated by using the input peak value of the F_v , and the average loading rate
198 between the 20% and 80% of the impact phase ($G_{v,ave}$), as:

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

199

200 **3. RESULTS**

201 The biomechanical model chosen in this work allows an accurate BCoM estimation in
 202 locomotion (Halvorsen, Eriksson, Gullstrand, Tinmark, & Nilsson, 2009), and its
 203 adoption in jumping shows an error comparable to the literature. Indeed, two
 204 validation indices were estimated during the flight phase of the jumps: AV_1 (m/s^2)
 205 index represents an estimation of the gravity constant acceleration (g), expected to be
 206 $9.81 m/s^2$, while AV_2 (m) index is defined as the root mean square error among the
 207 model estimated and matched ballistic centre of mass trajectory (Rabuffetti & Baroni,
 208 1999). Their overall mean values and s.d. are respectively AV_1 (m/s^2) = $-9.836 \pm$
 209 0.027 , AV_2 (m) = 0.003 ± 0.002 .

210 In Table 1 the results of all the experiments are shown. The visceral mass (VMD),
 211 pectoral and abdomen external mass displacements (EMD) are represented as relative
 212 to the BCoM. The VMD, for all the subjects, measured during normal jumps ($0.069 \pm$
 213 0.020 m), is significantly higher ($p < 0.05$, paired t-test), than in controlled jumps
 214 (0.053 ± 0.018 m). The average time courses of normal and controlled VMD are
 215 shown in Figure 4, while the mean individual curves of participants are displayed in
 216 Figure 5.

217 For all the subjects, VMD shows a different pattern with respect to the container
 218 displacement both in normal and in controlled jumps (Figure 4), with a detectable
 219 phase shift between the curves. A paired t-test shows no significant difference of time
 220 shift, both during the aerial (normal 50.6 ± 10.4 ms – controlled 49.3 ± 9.4 ms) and
 221 landing (normal 51.2 ± 14.4 ms - controlled 49.8 ± 8.8 ms) phases, confirming a

222 constant phase shift in both jumping techniques. A local maximum in visceral mass
223 displacement ($\dot{s}(t) = 0$) is detectable at about 40-45% of jump period (time between
224 two subsequent BCoM peaks) (Figure 4) and could be classified as a typical artefact
225 of the foot impact on the force platform (Bisseling & Hof, 2006). The pectoral and
226 abdominal EMD values show no significant difference in the two jumping techniques
227 (paired t-test), but the pectoral EMD is significantly larger ($p < 0.05$, paired t-test) than
228 the abdomen EMD in both techniques (Figure 6).

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

234

235 4. DISCUSSION

236 The aim of this investigation was to test the effect of a combined respiratory/jumping
237 strategy, properly designed for compacting viscera in the abdominal cavity, in
238 limiting the vertical viscera motion during vertical jumps. Applying a previously
239 developed method (A. Minetti & Belli, 1994), by concurrently using inverse and
240 direct dynamics, we revealed that such a strategy reduced the vertical excursion up to
241 30%, with potential increases of the overall stiffness of the human trunk/body.

242 The VMD mean value measured was comparable with the literature: few quantitative
243 analyses were conducted mostly anatomically (Beillas, Lafon, & Smith, 2009) or in
244 slow-dynamic condition (Hostettler, Nicolau, Remond, Marescaux, & Soler, 2010),
245 where vertical viscera motion was found to range between 0.03 m and 0.07 m. Only
246 Minetti & Belli reported a value related to submaximal repeated jumps (0.08 m),

247 while Boussuges and collaborators (Boussuges, Gole, & Blanc, 2009) set the limit of
248 vertical displacement on maximal diaphragm motion (0.070 ± 0.011 m).

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

258 In the literature several authors have already speculated about frequency and phase
259 coupling between respiratory and locomotory rhythms as affected by training
260 (Bernasconi & Kohl, 1993) or workload (Rassler & Kohl, 1996), but no one provided
261 evidences of voluntary control of internal body dynamics through specific respiration
262 techniques, synchronously performed with body CoM oscillations. Only McDermott
263 (McDermott, et al., 2003), by investigating the relationship between
264 locomotor/respiratory coupling and training level, found that expert runners were
265 particularly skilled in synching their coupling during speed changes. Therefore, from
266 the energetic point of view, these interactions should be controlled to avoid energy
267 losses resulting in some extra-mechanical work done by muscles, and the time delay
268 calculated between BCoM and VMD curves in this investigation, reinforces this
269 hypothesis. In fact, the ‘economy’ of bouncing locomotion, such as running or
270 skipping, could be influenced and the mechanical external work calculated from
271 kinematically measured CoM displacement could be refined by adding viscera

272 contribution (Daley & Usherwood, 2010). While this is supposed to be a small
273 adjustment in normal subjects, any deviation from a mesomorphic body such as obese
274 patients with relevant internal and external wobbling masses would involve a more
275 substantial correction of the inverse dynamics approach. In this way the proposed
276 respiratory strategy could give potential benefits in terms of movement performance
277 and the non-invasive method described could be easily adopted.

278 In terms of data processing the previous method (Minetti & Belli, 1994) has been
279 refined: kinematic sampling frequency has been quadrupled (400 Hz) and chosen as a
280 submultiple of the dynamometric signal to facilitate synchronization, the signals were
281 accurately aligned (double integration time gap), and the mathematical model was
282 validated with physics laboratory simulation software. Besides, the method still
283 suffered of inaccuracies due to: 1) the rigid body model assumption (Cappozzo, Della
284 Croce, Leardini, & Chiari, 2005; Chiari, Della Croce, Leardini, & Cappozzo, 2005)
285 originating troublesome theoretical interpretations of the results: the discrepancy
286 between the BCoM estimates from direct and inverse dynamics is considered as an
287 indirect evidence of viscera motion, but this could be partially the results of
288 experimental inaccuracies, 2) the “skin marker artefact” (Cappozzo, Catani, Leardini,
289 Benedetti, & Croce, 1996), which particularly affects movements with considerable
290 joint rotation as sit-to-stand (Kuo, et al., 2011) or locomotion (Akbarshahi, et al.,
291 2010) rather than vertical jumps with the arms blocked on the trunk, 3) the “soft tissue
292 motion artefact” (Gruber, et al., 1998; Leardini, et al., 2005), which can be assessed
293 by accelerometers (Kitazaki & Griffin, 1995) or by adding extra markers for the
294 oscillating body parts, at the cost of a more complex biomechanical model. The 4
295 'technical' markers introduced here, positioned on the estimated centre of mass of the
296 most visible and bulky 'external' wobbling masses (pectorals and abdominal muscles),

297 allowed their movement to contribute to refine VMD estimation. This simplified
298 approach does not completely compensate for the rigid body assumption inaccuracies
299 and cannot separate viscera from limbs soft tissues contribution (Gunther, et al.,
300 2003), but it constitutes an acceptable trade-off between ideal VMD estimation and
301 practical feasibility.

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

314 To date, soft tissues influences has already been investigated in locomotion (DeVita,
315 et al., 2007; Zelik & Kuo, 2010) and in jump landing (Gittoes, et al., 2006; M. T. Pain
316 & Challis, 2006), though its role still needs to be ultimately assessed. In this work,
317 even if there are several limitations, we compared two refined estimations of the most
318 influent soft tissue (viscera) motion in a simple motor task, repeatedly executed in the
319 same experimental condition. Indeed, subjects executed comparable jumps
320 considering the jumping frequency (f_{jump}), contact time (t_c), frequency of input force
321 (f_{GRF}) and the performance (body CoM vertical excursion). These evidences help to

322 minimize systematic and random errors, showing a de-noised measure of viscera
323 vertical excursion.

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

334

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338

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

490 Figure 2: (a) Mechanism used to generate an intra-abdominal pressure that compacts
491 the visceral mass: the subject after a combined deep diaphragmatic inspiration and
492 contraction of the abdominal “press” increases the intra-abdominal pressure also
493 executing progressive and short exhalations. The black arrows indicate: (1) The
494 lowering of the diaphragm that pushes on the viscera during inspiration (downward-
495 pointing white arrow); (2) The musculature of the abdominal “press”, which
496 contraction contributes to the elevation of intra-abdominal pressure (upward-pointing
497 white arrows). (b) On the left the overall mean (normalized in respect of the maximal
498 contraction value) and s.d of all the subjects, of rectus and obliquus abdominis muscle
499 activation, in normal (light-grey) and controlled (dark-grey) jump are shown. The
500 rectus and obliquus muscle activation is significantly higher in controlled jumps (* =
501 $p < 0.01$). On the right the overall mean and s.d., of the expired volume (V) during a
502 jump are shown. The expired volume is not significantly different between the
503 techniques.

504 Figure 3: Model used for the estimation of visceral mass displacement: M is the
505 container mass, m the internal visceral mass, and m_e is the external mass, while y_1 ,
506 y_2 and y_3 are distances from ground level and $s=y_2-y_1$. The whole system oscillates
507 vertically and exerts a vertical ground reaction force F_v , while internal and external
508 mass exerts a force f_v and f_e respectively on the container.

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

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

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

520 curve of aEMD (abdominal external mass displacement) in normal (grey solid line)
521 and controlled (grey dashed line) jumps, and the overall mean curve (controlled and
522 normal) of body CoM (black dotted line). All the curves are time-normalized with
523 single jump duration (0-100%). The pEMD and aEMD, for all the subjects, are not
524 significantly different in the two techniques, but the pEMD is significantly higher ($p <$
525 0.05) than aEMD both in normal and in controlled jumps.

526 Table 1: The mean and s.d. values of (1) visceral mass displacement (VMD), (2) body
527 CoM displacement (CoM), (3) pectorals (overall mean of right and left) external mas
528 displacement (pEMD), (4) abdomen (overall mean of right and left) external mass
529 displacement (EMD), (5) estimated input frequency (fGRF), (6) jumping frequency
530 (f_{jump}) and (7) contact time (t_c) in “normal” and “controlled” jumps are presented for
531 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	<i>S1</i>	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
	<i>S2</i>	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
	<i>S3</i>	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
	<i>S4</i>	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
	<i>S5</i>	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
	<i>S6</i>	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
	<i>All</i>	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	<i>S1</i>	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
	<i>S2</i>	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
	<i>S3</i>	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
	<i>S4</i>	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
	<i>S5</i>	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
	<i>S6</i>	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
	<i>All</i>	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
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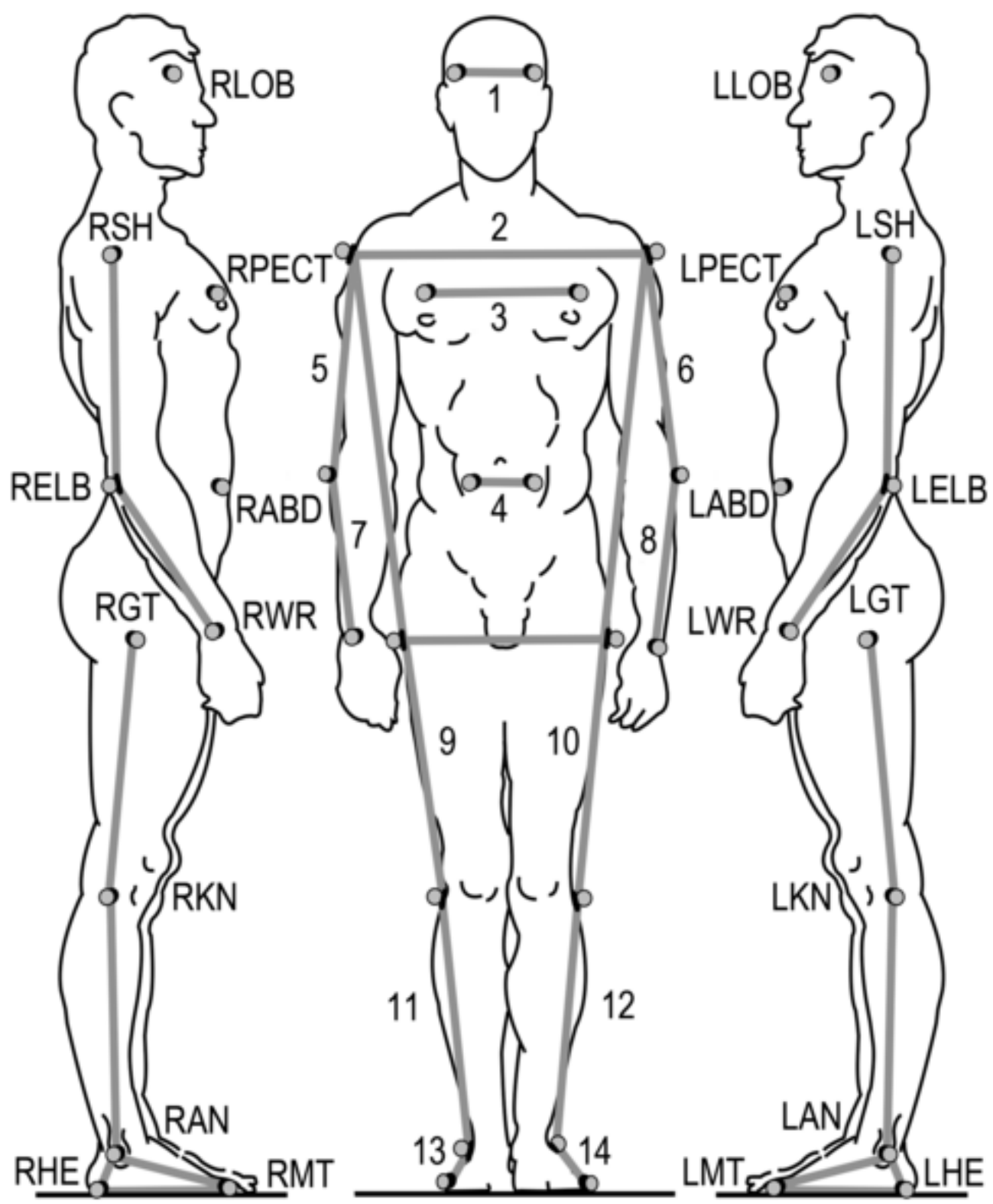


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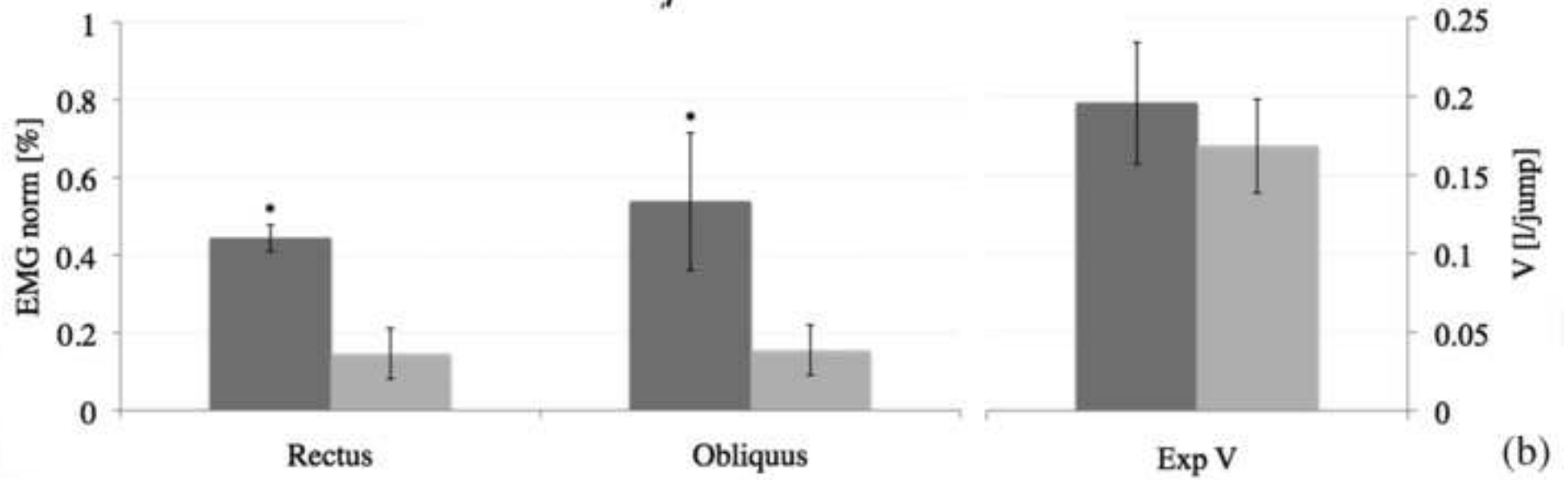
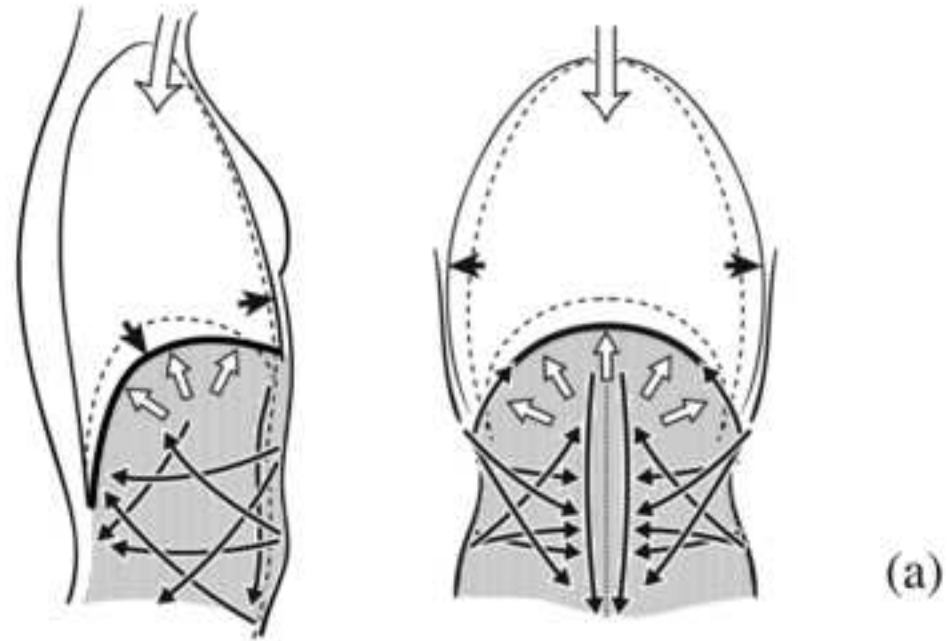


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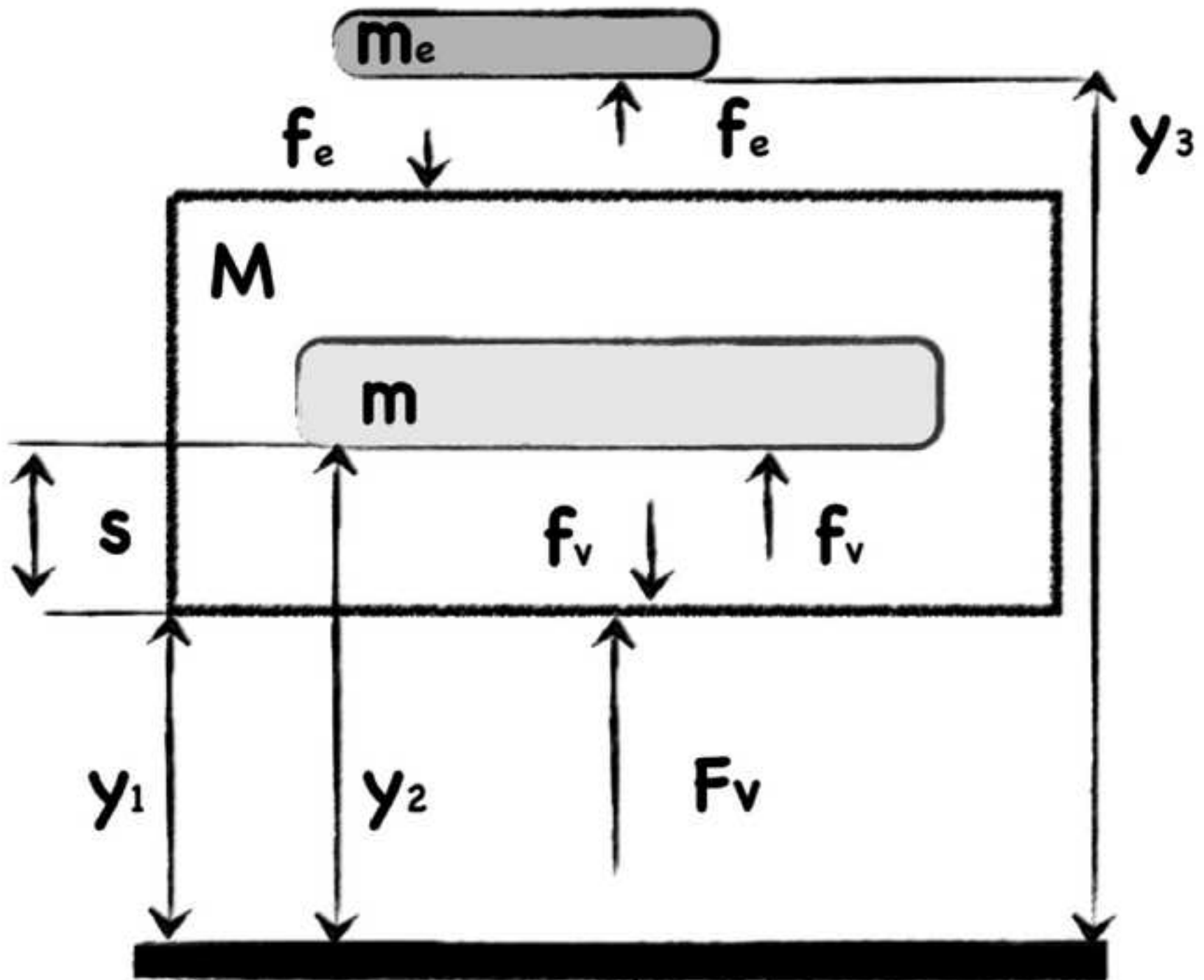


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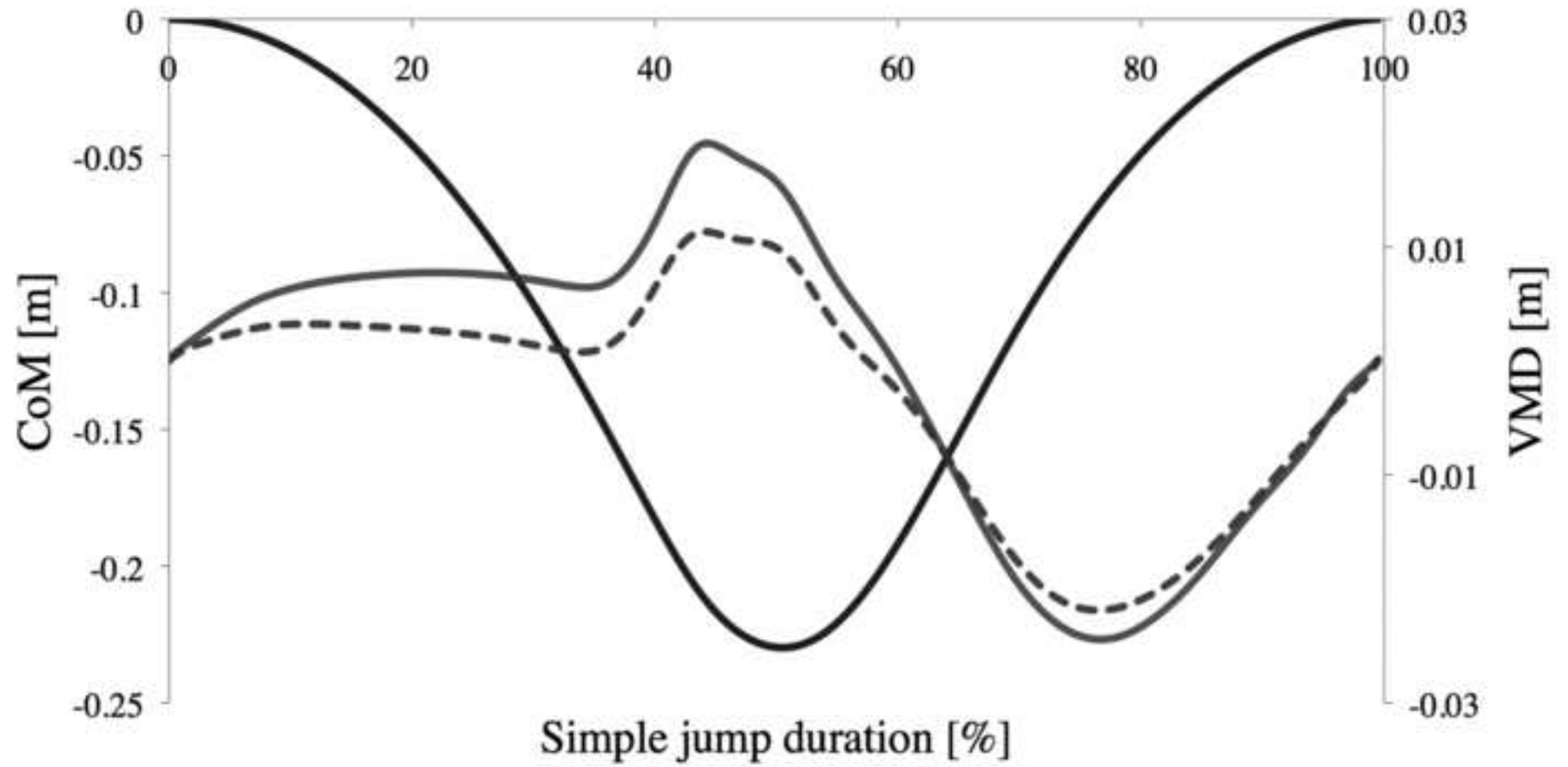


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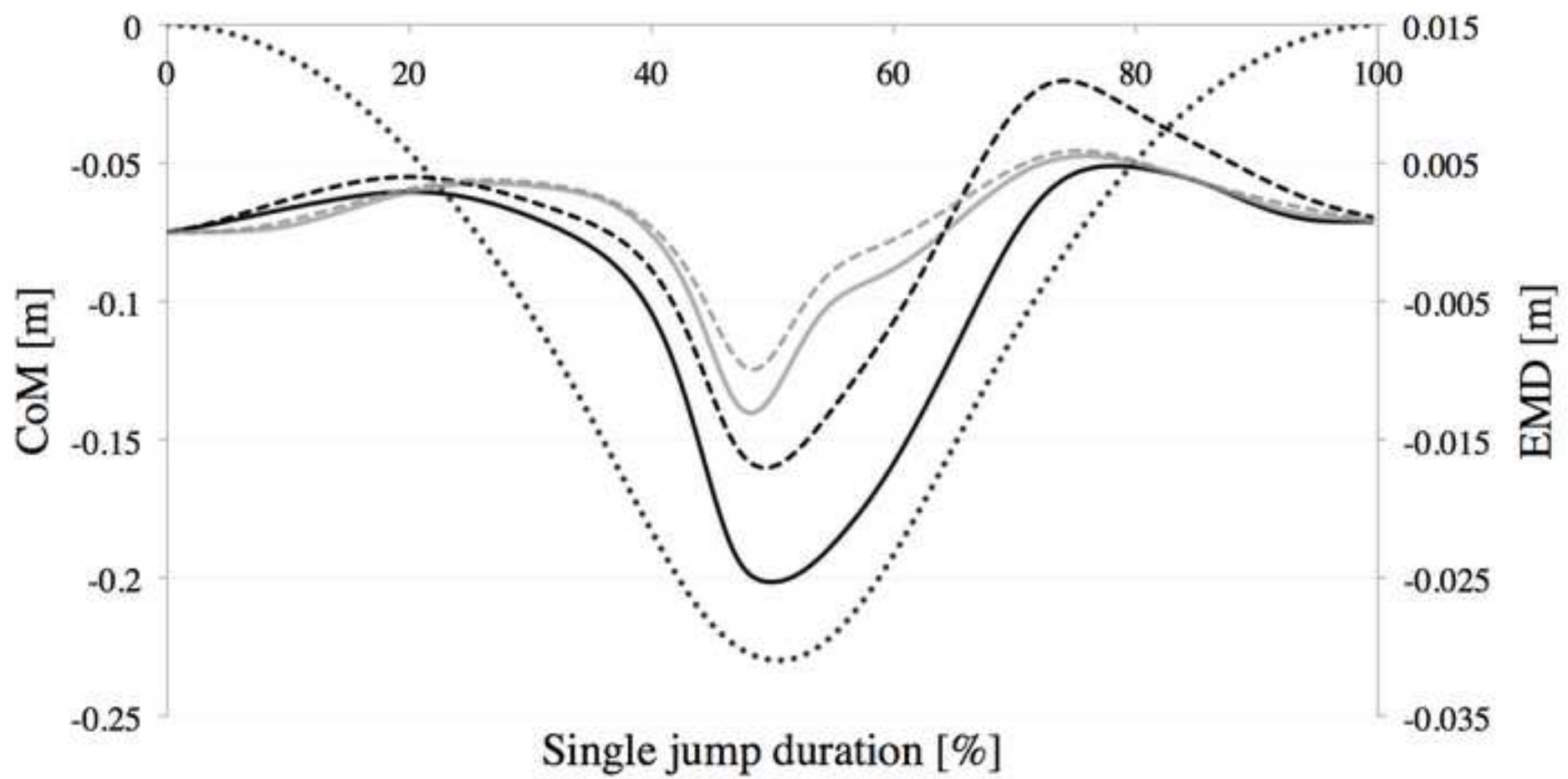


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