

1 **Simple models highlight differences in the walking biomechanics of**
2 **young children and adults**

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17 **Abstract (250 max)**

18 Adults conserve metabolic energy during walking by minimizing the step-to-step
19 transition work performed by the legs during double support and by utilizing spring-like
20 mechanisms in their legs, but little is known as to whether children utilize these same
21 mechanisms. To gain a better understanding, we studied how children (5-6 years) and adults
22 modulate the mechanical and metabolic demands of walking at their preferred speed, across slow
23 (75%), preferred (100%), and fast (125%) step frequencies. We quantified the 1) positive mass-
24 specific work done by the trailing leg during step-to-step transitions and 2) the leg's spring-like
25 behavior during single support. On average, children walked with a 36% greater net cost of
26 transport (COT; J/kg/m) than adults ($p=0.03$), yet both groups increased their net COT at varying
27 step frequencies. After scaling for speed, children generated ~2-fold less trailing limb positive
28 scaled mechanical work during the step-to-step transition ($p=0.02$). Unlike adults, children did
29 not modulate their trailing limb positive work to meet the demands of walking at 75% and 125%
30 of their preferred step frequency. In single support, young children operated their stance limb
31 with much greater compliance than adults ($\hat{k}= 6.23$ vs. 11.35 ; $p=.023$). Our observations suggest
32 that the mechanics of walking in children 5-6 years are fundamentally distinct from the
33 mechanics of walking in adults and may help to explain a child's higher net COT. These insights
34 have implications for the design of assistive devices for children and suggest that children cannot
35 be simply treated as scaled down versions of adults.

36 **Introduction**

37 Historically, simple models have been instrumental for understanding how humans
38 control the lifting and forward motion of the body's center of mass during walking. Simple
39 models reduce the body to a point mass supported by two stiff or, alternatively, spring-like struts
40 that characterize an "optimal" transfer or redirection of mechanical energy of the center of mass
41 (Antoniak et al., 2019; Cavagna, GA and Margaria, R, 1966; Donelan et al., 2002; Geyer et al.,
42 2006). With more efficient transfer or redirection, less mechanical work is required of the
43 muscles and tendons to propel the center of mass. Lowering the mechanical work required for
44 walking is associated with lowering the demand for consuming metabolic energy. This is now
45 recognized as a fundamental principle of locomotion biomechanics, and one might reason that
46 these simple models, for which these principles were derived from experimental data on adults,
47 could be applied to young children.

48 However, recent insights suggest that minimizing the work performed by the legs is a
49 dynamic process that must be learned, and which may depend on the biological and
50 morphological constraints of the body. Bril et al. (2015) use a simple model to show how young,
51 typically developing children (1 to 5 years) gradually learn to modulate the distance between
52 their center of mass and center of pressure, which dynamically changes during the gait cycle.
53 Modulating the distance between the center of mass and center of pressure is governed by the
54 forces that the legs generate in the vertical and anterior-posterior directions. But in particular, it
55 appears that generating forces in the anterior-poster direction (i.e. propulsive forces) is a skill
56 learned much later in childhood, which requires tuning and control to achieve balance and elicit
57 the desired motion in the forward direction (Brenière and Bril, 1998; Bril et al., 2015). Several
58 studies have also highlighted differences between the typical ground reaction profiles generated
59 by young children and adults (Dewolf et al., 2020; Takegami, et al., 1992), and have used simple
60 models to suggest mechanical work minimization may not be the optimal strategy for young
61 children (Usherwood et al., 2018b). This raises the question as to how the walking pattern of
62 young children should be modeled, recognizing that that their walking mechanics may not
63 simply be scaled down versions of adults.

64 For the same simple models to apply and scale to children, several assumptions must be
65 made about their walking behavior. First, we must assume that the inertial and gravitational

66 forces characteristic of walking will scale in proportion so that comparisons can be made across
67 “scaled speed.” Dimensionless speed, e.g. the Froude number, is a widely used method to scale
68 when comparing walking dynamics between children and adults; however, the Froude equation
69 is based on the idea of dynamic similarity (R.M. Alexander and A.S. Jayes, 1983). Dynamic
70 similarity means that multiplying all linear dimensions, time intervals, and forces by constant
71 factors would result in identical walking patterns. Deviations from dynamic similarity are
72 observed in young children (Kramer and Sarton-Miller, 2008; Usherwood et al., 2018b), and the
73 Froude number makes no allowance for differences in shape. Differences in shape, or
74 anthropomorphic proportions, may be an important consideration when making metabolic
75 comparisons within a species, or in animals relatively close in size (Kramer and Sylvester, 2013).
76 Alternative scaled speeds, like the Strouhal and Groucho (Alexander, 1989; Blickhan, 1989;
77 McMahon et al., 1985) may be used when elastic forces are considered, such as in running or
78 when a spring-like element is added to a walking template (Geyer et al., 2006).

79 Second, we must assume that young children have the muscular capacity to generate and
80 control the force and mechanical work required to move the center of mass in the most
81 economical way possible. During walking, the muscular capacity of muscles surrounding the hip
82 and the ankle joints are particularly important for propulsion and stability (Brenière and Brill,
83 1998; Ishikawa et al., 2005). Compared to adults, young children display many differences in
84 muscle and tendon properties. They have a proportionally smaller cross-sectional area of plantar
85 flexor muscles, differences in Achilles tendon compliance (Waugh et al., 2012), slower rates of
86 muscle force development, and lower magnitudes of relative maximum force production during
87 isolated contractions (Radnor et al., 2018). Further experiments would help to determine how
88 these properties influence functional or metabolic differences during walking. However, given
89 the importance of muscles and tendons to supporting body weight and redirecting the center of
90 mass – both critical for economic walking in adults (Donelan et al., 2002; Grabowski et al.,
91 2005) – it is conceivable that immature muscle-tendon capacities influence a young child’s
92 “optimal” walking solution. Of the few studies that specifically examine center of mass motion
93 in children, normalized center of mass amplitudes in the sagittal plane are proportionally greater
94 than in adults until around the age of 7 to 9 years (Dierick et al., 2004), which may reflect a
95 crucial period of dynamic changes in the maturation of both muscle and tendon (Malloggi et al.,
96 2019; Waugh et al., 2013).

97 Finally, we must assume that children minimize their metabolic cost during walking, and
98 that this cost is proportional to their size. Overall, when examining walking at a range of speeds,
99 children over 6 years exhibit an “optimal” speed that minimizes cost (DeJaeger et al., 2001). Yet,
100 surprisingly, at “optimal” speed, the net mass-specific cost of transport (COT, J/kg/m) is up to
101 33% higher in children less than 9 years old (Bolster et al., 2017; DeJaeger et al., 2001; Morgan
102 et al., 2002). With walking speeds normalized to the Froude number, DeJaeger et al. found that
103 differences in net COT may largely be reduced, thus body size alone may account for the higher
104 net COT observed in young children (DeJaeger et al., 2001). However, Schepens et al. found that
105 efficiency, defined as the ratio of the total mechanical power to the net energy consumption rate,
106 is much lower in younger children (Schepens, 2004). If mechanical power incurs a relatively
107 greater cost in younger children, this again raises the question as to the extent to which muscular
108 capacity and control plays a role in explaining higher net COT. Taken together, these insights
109 suggest a functional, mechanistic explanation for why young children are less economical at
110 walking than adults (Schepens, 2004).

111 To gain greater insight into a potential explanation, our objective was to use the
112 methodology of simple biomechanical templates to investigate how the generation of limb forces
113 and the resulting motion of the center of mass in children may differ from adults, and thus may
114 relate to their metabolic differences. We took a systematic approach by breaking down the
115 walking gait cycle into distinct phases of double and single support. In mature walking patterns,
116 the legs must redirect the center of mass from a downward and forward velocity to an upward
117 and forward velocity during double support, when both the leading and trailing limb are in
118 contact with the ground (Kuo et al., 2005). This redirection, also called the step-to-step
119 transition, is considered a major determinant of the metabolic cost of walking in adults (Donelan
120 et al., 2002), and has also been studied in 12-18 month old toddlers (Hallemans et al., 2004). For
121 our first aim, we sought to compare the positive external mechanical work (W_{ext}^+) performed by
122 the trailing leg in young children (age 5-6 years) and adults (age 18-30), with adults representing
123 the “ideal” behaviour. We hypothesized that 1) after accounting for differences in mass and
124 dimensionless walking speed, the trailing limb in young children would produce less W_{ext}^+
125 during double support. This hypothesis was based on literature that supports the idea that during
126 development, children gradually tune and increase their anterior-posterior propulsive forces (Bril
127 et al., 2015), indicating that when normalized to body mass, children aged 3-8 years generate less

128 power at the ankle than adults (Chester et al., 2006). Young children also typically produce
129 asymmetric vertical ground reaction forces while walking (Preis et al., 1997; Takegami, M.D.,
130 1992; Usherwood et al., 2018a), suggesting that children do not transition from one step to the
131 next like adults.

132 Inspired by the bipedal walking spring mass model of Geyer et al (2006), we also set out
133 to compare how young children and adults modulate the distance between their center of mass
134 and center of pressure during single support, which reflects the spring-like behavior of the leg.
135 From the perspective of a spring-mass model (Geyer et al. 2006), the redirection of the center of
136 mass can be achieved by the release of elastic energy that was stored during single support. At
137 the beginning of single support, when the leg extends and center of mass rises, elastic energy that
138 was absorbed during the double support may be released, and subsequently, as the center of mass
139 descends during the second half of single support, elastic energy can be stored in preparation for
140 double support (Donelan et al., 2002). This appears to be the ideal behavior in adults, whereby
141 step-to-step transitions are facilitated by the ability of the leg to store and release elastic energy
142 and thus, operate much like a spring. In contrast to adults, a child's leg consists of immature
143 muscles and tendons and may not operate like an ideal spring. In addition, single support is also
144 the phase that requires postural adjustment to stabilize the body in an upright position. The center
145 of mass is accelerated forward and sideways at the same time toward the swing leg (Bril et al.,
146 2015), having a destabilizing effect. Modulating the distance between the center of mass and
147 center of pressure, i.e. the spring length, during single support may be less precise in children
148 and require more muscle cocontraction and activation (Grosset et al., 2008; Lambertz et al.,
149 2003). These biomechanical constraints would ultimately raise the net COT during walking and
150 therefore, we hypothesized that 2) after considering dimensionless speed and scaling for size, the
151 stiffness of the leg spring (k) in children would differ from adults.

152 For both of our main hypotheses, we also explored whether young children would
153 modulate their trailing limb work and spring stiffness in the same way as adults when meeting
154 the demands of walking at a fixed speed, but at step frequencies slower and faster than preferred.
155 Therefore, we tested these hypotheses under conditions in which children and adults walked at
156 their preferred speed and across a range of slow to fast step frequencies set at 75%, 100%, and
157 125% of their preferred step frequency. Faster step frequencies involve shorter step lengths,

158 while slower step frequencies involve longer step lengths. At slower step frequencies, the angle
159 between the legs at the instance of double support increases and vice versa at faster step
160 frequencies. According to the individual limbs model, positive mechanical work at push-off
161 depends on both the center of mass velocity and on the angle between the legs (Donelan et al.,
162 2002). Following that model, we expected more propulsive trailing limb work at the slower step
163 frequency condition for both adults and children. In regards to the the bipedal spring mass
164 model, we expected that a slower step frequency would yield a decrease in the touchdown angle
165 of the leading limb; however, both touchdown angle and k vary within a large range (Geyer et
166 al., 2006), so we could not formulate a predictive hypothesis as to how k might change with step
167 frequency.

168 Overall, we reasoned that changes in the mechanical demands of modulating step
169 frequency would result in greater metabolic demands for young children. Specifically, when
170 challenged to walk at the slowest step frequency, which requires the largest step lengths, young
171 children would likely consume greater rates of metabolic energy, and hence a greater net COT
172 when compared with the other step frequency conditions. Given known differences in a child's
173 muscle and tendon morphology, ankle power generation, and lower walking efficiencies, we
174 hypothesized that when compared to adults, 3) young children would incur a greater net COT to
175 modulate their step frequency than adults. In particular, we expected that the legs would be
176 required to generate the greatest amount of mechanical work at the slowest step frequency
177 condition, and that the extra mechanical demand would be more costly for children than for
178 adults.

179

180

FIGURE 1 ABOUT HERE

181

182 **Materials and Methods**

183 Experimental Procedures

184 Healthy young adults (aged 18-32 years, $n=8$) and typically developing, healthy children
185 (aged 5-6 years, $n=8$) were recruited locally. Adult subjects gave their written informed consent
186 to participate. Parents of child subjects gave permission and written informed consent, and

187 children gave verbal assent to participate, in accordance with ethical guidelines and approved by
188 University of Houston Institutional Review Board. To ensure that we captured the most natural
189 walking patterns and comfort with the testing environment, children visited the lab for a
190 preliminary 1.5-hour practice session the previous day. For the child cohort, a parent was present
191 during all parts of the experiment and gave encouragement when necessary.

192 During the day of the experiment, subjects arrived having fasted and refrained from
193 caffeine or exercise for at least 3 hours. Upon arrival to the lab, subjects rested for 10 min before
194 we measured their standing metabolic rate for 5 minutes using an open circuit TrueOne 2400
195 metabolic system (ParvoMedics, Inc. Sandy, UT USA). The metabolic system was calibrated
196 immediately before each session using standard gases and a 3L syringe. Following the standing
197 trial, reflective markers (15.9 mm) were placed according to manufacturer guidelines (Lower
198 Body Plug-in Gait, 100 Hz; Vicon 12-camera system, Nexus 1.8.5, Vicon, Oxford, UK) and the
199 standard scaling and calibration protocol of the Nexus 1.8.5 software was followed. An
200 additional marker was placed at the subject-specific location of the center of mass as determined
201 by the reaction board method (Enoka, 2015). All subjects (Table 1) wore their own shirt, bike
202 shorts, and tennis shoes with a heel-sole difference no greater than 6.4 mm.

203 Each subject then walked on a level dual belt instrumented treadmill (1000Hz; Bertec Co.
204 Columbus, OH USA) at their preferred speed. Preferred speed was obtained through feedback by
205 having subjects walk at increasing increments of speed (starting from 0.3 m/s in the child group
206 and 0.5 m/s in the adult group). Once they reached a preferred walking speed, speed was
207 increased again and lowered if necessary, to confirm their preferred walking speed (Arellano et
208 al., 2009). Following a 5 min rest, subjects then walked for 5 minutes at preferred walking speed
209 and were instructed to match their step frequency to the sound of a metronome at 75, 100, and
210 125% of preferred step frequency (order randomized). Subjects sat for a 5 min rest period
211 between trials. All subjects achieved a steady rate of metabolic energy consumption with
212 respiratory exchange ratios (RER) remaining within the normal physiological range below 1.0.

213 Data analysis

214 Walking metabolic power (W/kg) was calculated from average $\dot{V}O_2$ and $\dot{V}CO_2$ during the last
215 3 minutes of each trial (Brockway, 1987). The average quiet standing value was subtracted from
216 the average walking value to yield net metabolic power (W/kg). Net metabolic power was

217 divided by speed (m/s) to obtain net COT expressed in units of J/kg/m. Data were filtered using a
218 4th order, zero-lag low-pass Butterworth filter with a cut-off frequency of 15Hz for force and
219 6Hz for kinematics. All data were processed in Matlab (R2018b, The MathWorks,
220 Massachusetts, USA) and custom code was written to calculate values for limb work and power
221 (Donelan et al., 2002) and for values of leg stiffness, k , following a spring-mass model (Geyer et
222 al., 2006). To identify gait cycle events and periods of double and single support, we defined
223 touchdown and toe-off as the instant when the vertical ground reaction force (GRF) crossed a
224 threshold of 5% body weight. Double support was then defined as the portion of the gait cycle
225 after touchdown of the leading limb, and before toe-off of the trailing limb. Single support was
226 defined as the portion of the gait cycle between toe-off and touchdown when only one foot was
227 in contact with the ground.

228 *Individual Limbs Method.* Starting from the 3 min mark of each 5 min walking trial, filtered
229 force data from periods of double support were aggregated to determine lead leg and trail leg
230 power and work. Following the method of Donelan et al., the velocity of the center of mass was
231 determined by single integration. Then the external mechanical power generated by the trailing
232 and leading limb (as shown in Fig. 2) was determined by summing together the dot product of
233 the force and velocity of the center of mass acting in each direction (Donelan et al., 2002).

234 (1a) $P_{lead} = Force_{x,lead} \cdot velocity_{x,com} + Force_{y,lead} \cdot velocity_{y,com} + Force_{z,lead} \cdot$
235 $velocity_{z,com}$

236 (1b) $P_{trail} = Force_{x,trial} \cdot velocity_{x,com} + Force_{y,trial} \cdot velocity_{y,com} + Force_{z,trial} \cdot$
237 $velocity_{z,com}$

238 The magnitude of positive mechanical work of the trailing limb (W_{trail}^+), expressed in Joules, was
239 then calculated as the time integral of the positive portions of trailing limb power from equation
240 (1b). For comparisons of young children and adults at their preferred speed (m/s), the total W_{trail}^+
241 was normalized to mass (Schepens, 2004) and averaged across the first 100 double supports
242 starting from the 3 min mark.

243

244

FIGURE 2 ABOUT HERE

245

246 *Single Support Spring Mass Method.* Again, starting from the 3 min mark of each 5 min
247 walking trial, filtered force data from periods of single support were aggregated. We first found
248 the position of the center of mass using double integration and then defined a vector from the
249 center of mass to the foot's centre of pressure to represent a virtual 2D spring in the sagittal
250 plane. The markers at the center of mass and pelvis were used as an offset to reflect the absolute
251 position of the center of mass after double integration, such that the center of mass was assumed
252 to be within the body and half the distance between the anterior superior iliac spine and posterior
253 superior iliac spine markers. Following the work of Gard et al. (2004), we assume that this
254 method is a close approximation to the exact location of the center of mass and more
255 importantly, should reflect the trajectory and amplitudes that the center of mass undergoes during
256 each step. Since we were primarily interested in the lift and propulsion of the center of mass, we
257 reasoned that a sagittal spring was an adequate starting point for comparisons. Plotting resultant
258 2D GRF (N) values as a function of spring length (m) allowed us to calculate the slope of the
259 best fit line via a least-squares regression analysis (Fig. 3). The mean of the absolute slopes
260 estimated from the first 30 steps was reported as the value of k . This decision was based on our
261 finding that adults exhibited consistent spring-like behavior that revealed best fit line R^2 values
262 ranging from 0.5 to 0.9. On the other hand, children exhibited spring-like behavior that was
263 much less consistent, so for fair comparisons, we limited the R^2 values to 0.5 which allowed us
264 to aggregate a minimum of 30 steps.

265

266 **FIGURE 3 ABOUT HERE**

267

268 *Scaled speeds and Touchdown Angle.* The Froude number, as shown in Eq. 2, normalizes
269 speed based on pendular dynamics (R.M. Alexander and A.S. Jayes, 1983). We defined velocity,
270 v , as the speed of the treadmill, and measured leg length as the distance between the anterior
271 superior iliac spine to the distal tip of the medial malleolus. The dynamic walking model
272 equations only apply when the body conforms to inverted pendulum dynamics (Kuo et al., 2005),
273 so we chose to consider Froude speed in our comparison of individual limb work.

274
$$(2) Fr = \frac{v^2}{gL_{leg}}$$

275 To normalize k , we used the Groucho speed, which was originally developed as a vertical speed
276 parameter, combining interactions between effective vertical spring stiffness, gravity, impact

277 velocity, and body mass (McMahon et al., 1985). It has been mathematically adapted and used in
278 2 dimensions by also considering leg length (Blickhan, 1989). Essentially, for both the fore-aft
279 and vertical directions, the Groucho speed can be calculated using equation (3a) and normalized
280 to leg length, where v is the resultant 2D velocity at the instant of touchdown, g is gravity, and ω
281 is the natural frequency of the system as determined by spring stiffness, k , and body mass, m , as
282 in equation (3b).

$$283 \quad (3a) \quad Gr = \frac{v\omega}{g}$$

$$284 \quad (3b) \quad \omega = \sqrt{k/m}$$

285 Because the Groucho number is a dimensionless speed based on a spring-mass model, we
286 planned to use Groucho speed as a covariate in our comparison of k , but as explained in the
287 Statistical analysis below, we instead used \hat{k} , which represents its non-dimensional form.
288 Touchdown angle was determined in the sagittal plane as the angle formed at touchdown
289 between a line created by the center of mass and lateral malleolus marker with respect to the
290 horizontal. We aggregated the touchdown angles from the entire 3 minutes and report the means
291 in Table 2.

292

293 Statistical analysis

294 We used separate mixed R-ANCOVA's with *a priori* planned comparisons to test for
295 differences in W_{trail}^+ , non-dimensional spring stiffness (\hat{k}), touchdown angle, and net COT. After
296 checking for normality, we compared differences between groups at the 3 step frequencies by
297 defining age and step frequency as a between and within subjects fixed factor, respectively. To
298 account for differences in size and speed between children and young adults, we included in our
299 analysis of W_{trail}^+ , the dimensionless speed Froude as a covariate.

300 Upon initial inspection of our Groucho speeds, we removed an outlier in the adult group so
301 statistical comparisons for k were based on $n=7$ in the adult group. Based on independent *t*-test at
302 each step frequency (Table 2), we did not detect significant differences in Groucho speed
303 between groups and therefore did not require Groucho as a covariate. We did not use Froude as a
304 covariate for this analysis, since we were only concerned with single support and were using a
305 theoretical model that is based on spring-mass dynamics, not pendular dynamics. To account for

306 differences due to body size in our comparison of k , we transformed to non-dimensional \hat{k} using
307 equation 4, where l is the max spring length, m is body mass, and g is gravity.

308
$$(4) \hat{k} = \frac{kl}{mg}$$

309 For comparisons of touchdown angle and net COT (J/kg/m), we included Froude as a
310 covariate because these variables strongly depend on walking speed. During post-processing, we
311 discovered that a hardware malfunction caused the force data to be erased for one child subject
312 walking at 125% of their preferred step frequency. Therefore, statistical comparisons for mass-
313 normalized W_{trail}^+ and \hat{k} were based on a sample size of $n=7$ in the child group. For our planned
314 comparisons, we used independent t -tests when normality was met or Mann-Whitney U tests
315 when normality was not met. Finally, we plotted mass-normalized W_{trail}^+ , \hat{k} , and net COT across
316 conditions for both groups. All data was analysed in SPSS with statistical significance set at 0.05
317 (version 26, IBM, Armonk, NY).

318

319 **Results**

320 *Touchdown angle.* We used Froude as a covariate to compare touchdown angle between
321 groups and confirmed that touchdown angle, an indirect gauge for detecting differences in step
322 length, was changing with step frequency as expected. A significant main effect for step
323 frequency ($F_{2,24}=3.324$, $p=.027$, $\eta^2=.217$) revealed that touchdown angle increased on average
324 from ~72 to 78 degrees when walking from the relatively slow to fast step frequencies (Table 2).
325 We did not detect a main effect for group or an interaction effect between group and step
326 frequency, indicating that in response to the changing step frequencies, children and adults
327 altered their touchdown angle in the same manner.

328 *Double Support Mechanics.* Prior to conducting the statistical analysis, data were inspected
329 to ensure assumptions were met for a repeated measures-ANCOVA, with Froude speed as an
330 appropriate covariate. We detected a significant within-subjects difference of W_{trail}^+ /kg across
331 step frequencies ($F_{2,24}=3.736$, $p=.039$, $\eta^2=.237$). As expected in adults, overall positive
332 propulsive work was greater at longer step lengths (75% preferred step frequency) and less at
333 shorter step lengths (125% preferred step frequency; Figure 4A). This was not so in the child
334 group, as confirmed by our detection of an interaction effect ($F_{2,24}=5.216$, $p=.013$, $\eta^2=.303$),

335 indicating that children did not modulate the amount of $W^+_{\text{trail}}/\text{kg}$ when walking at longer and
336 shorter steps associated with the 75% and 125% step frequency condition, respectively (Figure
337 4A). Between group comparisons across step frequency conditions confirmed that after
338 accounting for differences in Froude speed, children and adults generated 0.06 J/kg and 0.12 J/kg
339 of trailing limb mass-specific positive work, respectively ($F_{1,12} = 14.106$, $p = .002$, $\eta^2 = .54$). Pre-
340 planned ANCOVA comparisons between groups confirmed that children generated less $W^+_{\text{trail}}/\text{kg}$
341 than adults at each step frequency condition (Table 2, all $p < .05$).

342 *Single Support Spring Mass Mechanics.* After checking that the statistical assumptions were
343 met for a repeated measures ANOVA, we detected a main effect for group ($F_{1,12} = 6.831$, $p < .001$,
344 $\eta^2 = .363$) and step frequency ($F_{2,24} = 46.43$, $p < .001$, $\eta^2 = .795$), but no interaction effect ($p = 0.41$).
345 At the slow step frequency associated with longer steps, \hat{k} decreased, and at the fast step
346 frequency associated with shorter steps, \hat{k} increased (Fig. 4B). Overall, children walked with an
347 average \hat{k} equal to 6.82, almost 2-fold more compliant than adults, who walked with an average
348 \hat{k} equal to 11.35. The much more compliant \hat{k} observed in children was consistent at each step
349 frequency condition (all p 's $< .05$ for all *a priori* independent *t*-tests; Table 2).

350 *Net Metabolic Cost of Transport.* Prior to conducting the analysis, data were inspected to
351 ensure assumptions were met for a mixed design and that the Froude number was an appropriate
352 covariate. A main effect for step frequency ($F_{2,28} = 19.656$, $p < .001$, $\eta^2 = .584$) revealed a higher
353 net COT when walking at step frequencies above and below preferred. We also detected a main
354 effect for group ($F_{1,14} = 31.84$, $p < .001$, $\eta^2 = .695$), revealing that the net COT was on average, 36%
355 higher in children (Fig. 4C). However, we did not detect an interaction effect ($p < 0.05$),
356 indicating that across slow and fast step frequencies, the net COT changed in a similar way for
357 adults and children. And finally, when adjusted for group differences in the Froude number, the
358 mean values show that in both children and adults, the net COT increased when walking at the
359 relatively slow and fast step frequencies.

360

361

FIGURE 4 ABOUT HERE

362

363 **Discussion**

364 We analyzed experimental data using simple models of walking to compare center of
365 mass mechanics and net COT in a group of young children to those of adults, who represent the
366 ideal behavior these simple models are based upon. One key finding was that trailing limb
367 positive work, W_{trail}^+ /kg, was significantly lower across conditions in children. This supported
368 our first hypothesis. However, in contrast to the other variables tested, the magnitude of this
369 difference changed depending on the step frequency condition. The largest deficit was evident at
370 the 75% step frequency—the condition that required subjects to take the longest steps. From
371 Donelan et al., (2002) and Kuo et al. (2005), it is predicted that longer steps will increase the
372 collision of the lead leg with the ground, effectively increasing negative work generated by the
373 leading limb. To account for this greater negative work, the trailing limb must generate more
374 positive work. Yet children did not generate the positive work that would be expected to account
375 for the increased collision forces that are associated with taking long steps at a fixed speed.

376 When walking at the preferred step frequency condition, we found that on average, the
377 trailing limb in children is used to both absorb and generate work, while in adults, the trailing
378 limb only generates positive work, providing almost 100% of all their propulsive positive work
379 (e.g., see Fig. 2, middle column). In adults, there seems to be a clear role for the trailing limb to
380 generate positive work and the leading limb to generate negative work, whereas young children
381 used both limbs to generate both positive and negative work. Halleman's et al. (2004) reports
382 that in toddlers (12-18 months), an inefficient inverted pendular mechanism of energy exchange
383 contributes to differences in external mechanical work. Toddlers are described as utilizing a
384 “tossing gait” where work that is performed to lift the center of mass against gravity is much
385 greater than work that is performed to propel the center of mass. In our age group of 5-6 years,
386 gait patterns are considered more mature than in toddlers. Yet, when partitioning the positive and
387 negative work generated by each limb double support, we also found (as in Hallemans et al.) that
388 our child subjects tended to generate more mass-specific work to lift the center of mass than in
389 adults. However, one key difference was that the mass-specific work to lift versus to propel the
390 center of mass was approximately equal for our child subjects (Fig. 5), suggesting a possible
391 shift with age toward more mature patterns when work to propel the center of mass dominates.

392

393

FIGURE 5 ABOUT HERE

394

395 In single support, we found that after scaling for size and speed, \hat{k} is approximately 40%
396 lower in our child group, which supported our second hypothesis. While \hat{k} was substantially
397 lower in our child group, they did modulate \hat{k} in the same way as adults. As shown in Fig. 4B,
398 both young children and adults increased \hat{k} when meeting the mechanical demands of walking at
399 fixed speed, but at relatively slower and faster step frequencies. The tendency of children to
400 modulate \hat{k} was independent of changes in the amount of positive work generated by their
401 trailing limb during double support. This behavior deviated from that observed in adults, where
402 the positive work generated by the trailing limb during double support decreased and \hat{k} increased
403 across slow to fast step frequencies. Unlike trailing limb work, which has been shown to rely
404 primarily on plantar flexor muscles and tendons (Fukunaga et al., 2001; Ishikawa et al., 2005;
405 Sawicki and Khan, 2016), \hat{k} can be influenced by multiple muscles and joints, such as the ankle,
406 knee, and hip. Thus, it is possible that both the more compliant \hat{k} and inability to generate as
407 much scaled work and power by the trailing limb may be related to a young child's developing
408 muscular capacity. Over time, as children learn to coordinate their leg and hip muscles to
409 generate propulsive work, the motion of the center of mass during walking may become
410 smoother, reflecting patterns that are observed in adults. Given evidence of structural differences
411 of immature plantar flexor muscle and tendon (Radnor et al., 2018; Waugh et al., 2012) and
412 functional differences in muscle and tendon interaction in children measured directly by
413 ultrasound while hopping (Waugh et al., 2017), it seems likely that the coordination of the
414 plantar flexor muscle-tendon mechanics responsible for efficient propulsion and redirection of
415 the center of mass in adults is not fully formed and learned at 5-6 years of age.

416 The unequal contribution of work by the trailing limb in young children could also help
417 explain their decreased efficiency (Schepens et al 2004), since positive work is necessary to
418 restore the energy that is lost from the unavoidable collision phase of double support (Donelan et
419 al., 2002). Lost energy must then be replaced with more costly compensation strategies, possibly
420 at other joints such as the hip (Lye et al., 2016; Sawicki et al., 2009), and with coactivation of
421 other muscles (Lambertz et al., 2003). This would ultimately raise the net COT, which is what
422 we observed in our 5–6-year-old child group. Consistent with previous studies (DeJaeger et al.,
423 2001; Morgan et al., 2002), our children had a higher net COT at each step frequency condition,

424 even after accounting for differences in Froude speed. When compared to adults, however, the
425 net COT in children did not fluctuate much in response to changes in step frequency (Figure 4C).
426 Thus, we reject our third hypothesis that when compared to adults, walking at the lowest step
427 frequency condition would be relatively more costly for children.

428 During the first 7-13 years of life, walking patterns vary in their maturity, so
429 interpretation of our results is limited to children aged 5-6 years. At this age, we observed a high
430 amount of variability in the ground reaction force patterns, consistent with previous studies
431 (Kraan et al., 2017). This variability ultimately meant that we were only able to reliably take 30
432 steps from each trial that met our condition of a reasonable linear fit between force and spring
433 length changes during single support. It is possible that the variability in ground reaction force
434 patterns might have another purpose, such as for motor learning, where the goal is to learn how
435 to efficiently redirect the center of mass. Indeed, the complex relationships between age, size,
436 mechanics, and the spatiotemporal components of walking remain to be understood, and we lack
437 a complete understanding of their contributions to the higher net COT observed in children. Our
438 results should be interpreted with caution because the sample size was small, though it is worth
439 noting that our effect sizes (Cohen's d) are between 1.07-10 (Table 2), which may be helpful for
440 future reference.

441 In summary, when compared to adults, we found that when walking at a fixed speed, but
442 at relatively slow and fast step frequencies, 5–6-year-old children generated significantly less
443 positive work by their trailing limb during double support. We also found that \hat{k} , the spring-like
444 behavior of the leg, was much more compliant in children than adults. These variables were
445 scaled to speed, mass, and leg length, yet the mechanics of walking in children departed
446 substantially from adults, who have long been seen as the ideal behavior for economical walking.
447 Altogether, our findings suggest that the simple models used here cannot be scaled down and
448 used to adequately characterize a child's walking mechanics and energetics. The implications
449 from this work are worth further consideration because simple models form the basis for more
450 complex theoretical and computational models and aid in the design of assistive devices such as
451 lower limb prostheses and orthoses (Delp et al., 2007; Gard and Childress, 2001; Geyer and
452 Herr, 2010). As opposed to relying on models that are scaled down versions of adults, we
453 propose that a limb-level biomechanical analysis of walking in children may facilitate the

454 selection and tuning of assistive devices that are specifically designed for children. It is well
455 known that adults can successfully tune and utilize the Achilles tendon to store and release
456 energy to generate propulsive work (Fukunaga et al., 2001; Ishikawa et al., 2005; Lichtwark and
457 Wilson, 2008), directing the attention of engineers and clinicians to the ankle as a first line target
458 for assistive technology in gait rehabilitation. However, little is understood about plantar flexor
459 muscle-tendon function in children, who also require this technology and, in our study, did not
460 generate comparable positive propulsive work by their trailing limb. Combining a limb-level
461 biomechanical analyses with ultrasound measurements of plantar flexor muscle-tendon function
462 is of future interest, as this would help understand the unique strategies that children use to meet
463 the mechanical power demands of walking across a range of speeds and uphill/downhill
464 conditions.

465

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471

472 **Competing interests**

473 No competing interests declared.

474

475 **Author Contributions**

476 Conceptualization: V.L.R., C.J.A.; Methodology: V.L.R., C.J.A.; Software: V.L.R., C.J.A.;
477 Validation: V.L.R., C.J.A.; Formal analysis: V.L.R., C.J.A.; Investigation: V.L.R.; Resources:
478 C.J.A.; Data curation: V.L.R., C.J.A.; Writing - original draft: V.L.R.; Writing - review &
479 editing: V.L.R., C.J.A.; Visualization: V.L.R., C.J.A.; Supervision: C.J.A.; Project
480 administration: C.J.A.; Funding acquisition: C.J.A.

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482

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485

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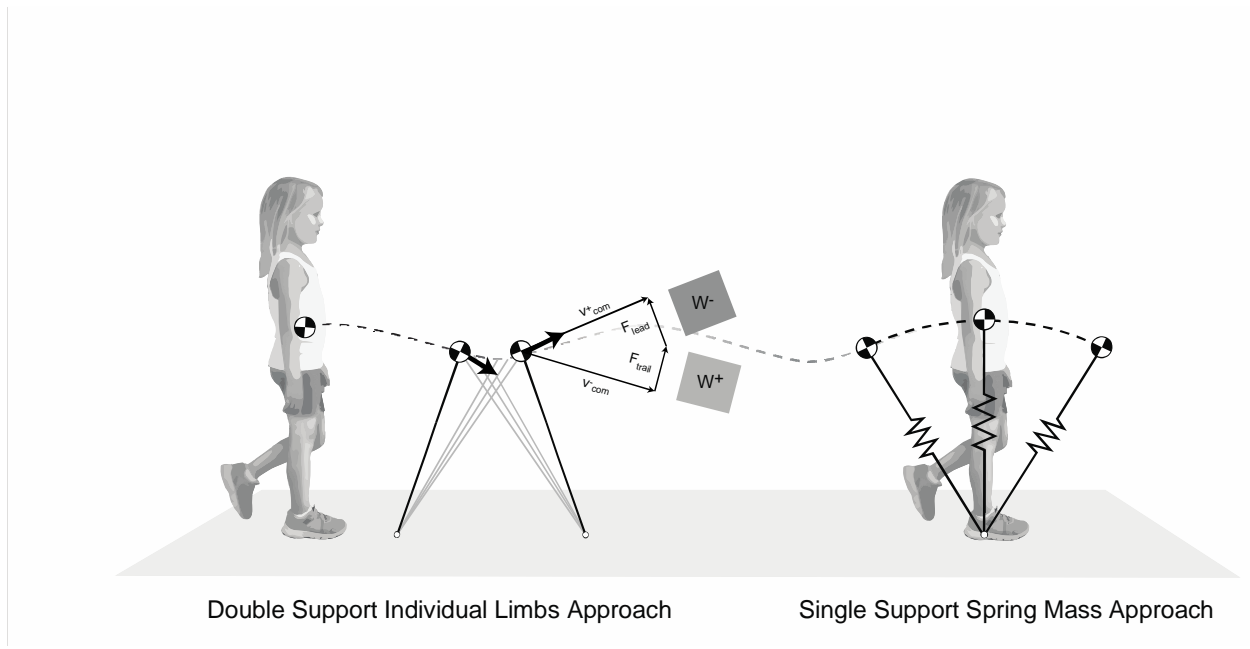
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606 muscle–tendon behaviour: implications for movement efficiency. *The Journal of*
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613 **Figure 1.** Geometric diagrams illustrate the templates used to analyze the mechanics of walking in
614 children. For both templates, ground reaction force data were used to calculate center of mass
615 and position. Following the dynamic walking model, (left) we quantified the mechanical work generated
616 by the individual limbs during double support (Donelan et al. 2002), a key determinant in transitioning
617 the body's center of mass from the trailing leg to the lead leg. In recognizing the contributions of elastic
618 energy storage and return to the work done on the center of mass, we quantified the spring stiffness, k , of
619 the leg as proposed by the spring-mass model (Geyer et al). As a simple approach, we quantified the
620 spring-like behavior of the leg during single support, as this is the period when the leg "spring" would
621 undergo energy release as the spring extends and the center of mass reaches it maximum height at
622 midstance. Then, as the center of mass moves forward and its height decreases, the spring compresses and
623 stores energy. The diagram depicting the step-to-step transition during double support is modified after
624 Kuo et al. (2005).

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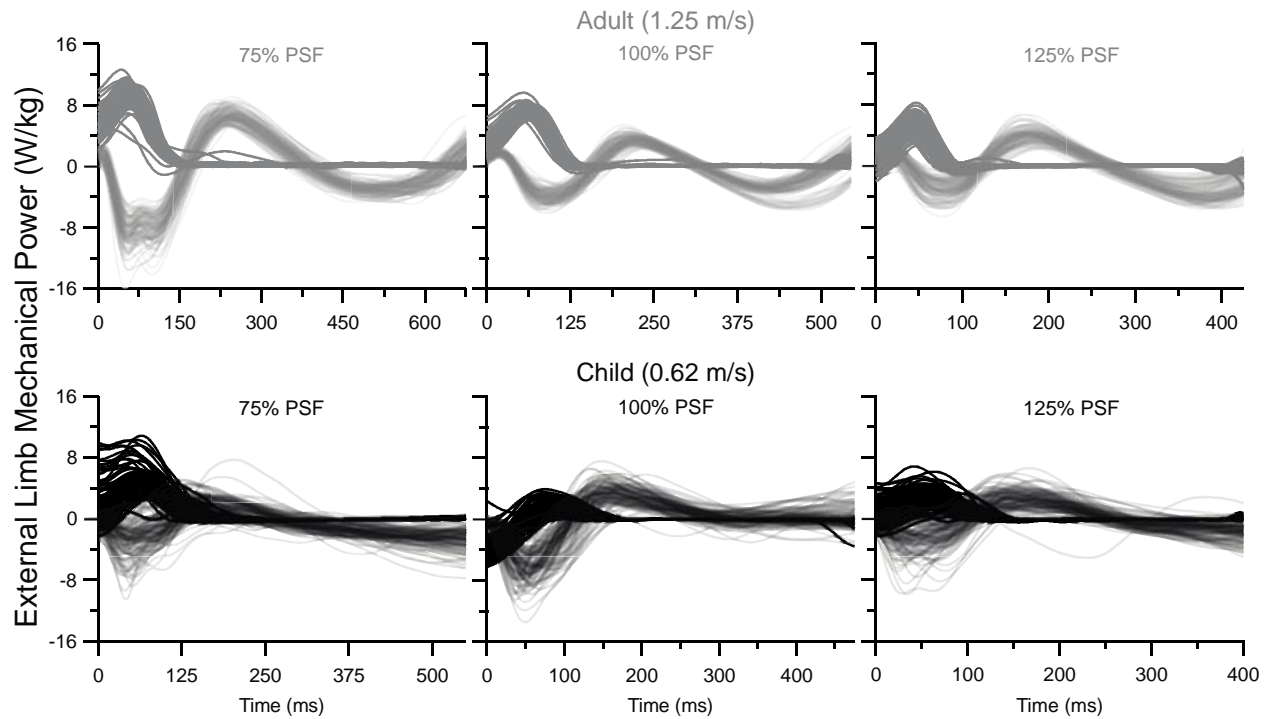
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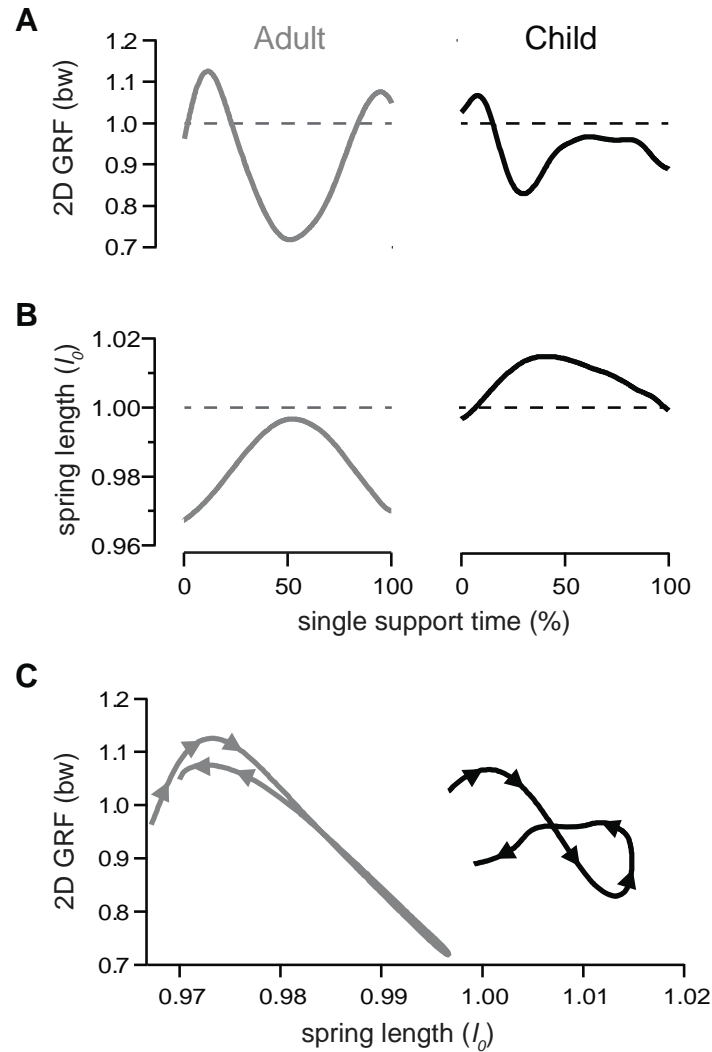
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638 **Figure 2.** Examples of an adult (gray lines, top) and child (black lines, bottom) external mechanical
639 power generated by the trailing (dark) and leading (light) leg during consecutive steps, under conditions
640 of walking at their self-selected speed at 75%, 100%, and 125% of preferred step frequency (PSF). When
641 walking at 1.25 m/s and at 100% PSF, the external mechanical power generated by the legs follows the
642 typical pattern observed in young, healthy adults (Donelan et al. 2002), with the trailing and leading leg
643 generating roughly equal positive and negative power during double support. In contrast, when walking at
644 their self-selected speed of 0.62 m/s at 100% PSF, the child's leading and trailing legs transitioned from
645 generating negative to positive external mechanical power during double support. In general, when
646 walking at a fixed speed, but at variable step frequencies, the adult modulated the magnitude of the
647 positive external mechanical power generated by their trailing leg, with the slower step frequency
648 coinciding with higher positive power and the faster step frequency coinciding with lower negative
649 power. The child, however, exhibited a strategy whereby the trailing leg no longer generates negative and
650 positive power during double support (as observed at 100% PSF), but primarily generates positive power
651 when walking at the slower and faster step frequency. To highlight the greater amount of variability
652 observed in our child group, plots of external mechanical power as a function of time consist of 100
653 consecutive steps.



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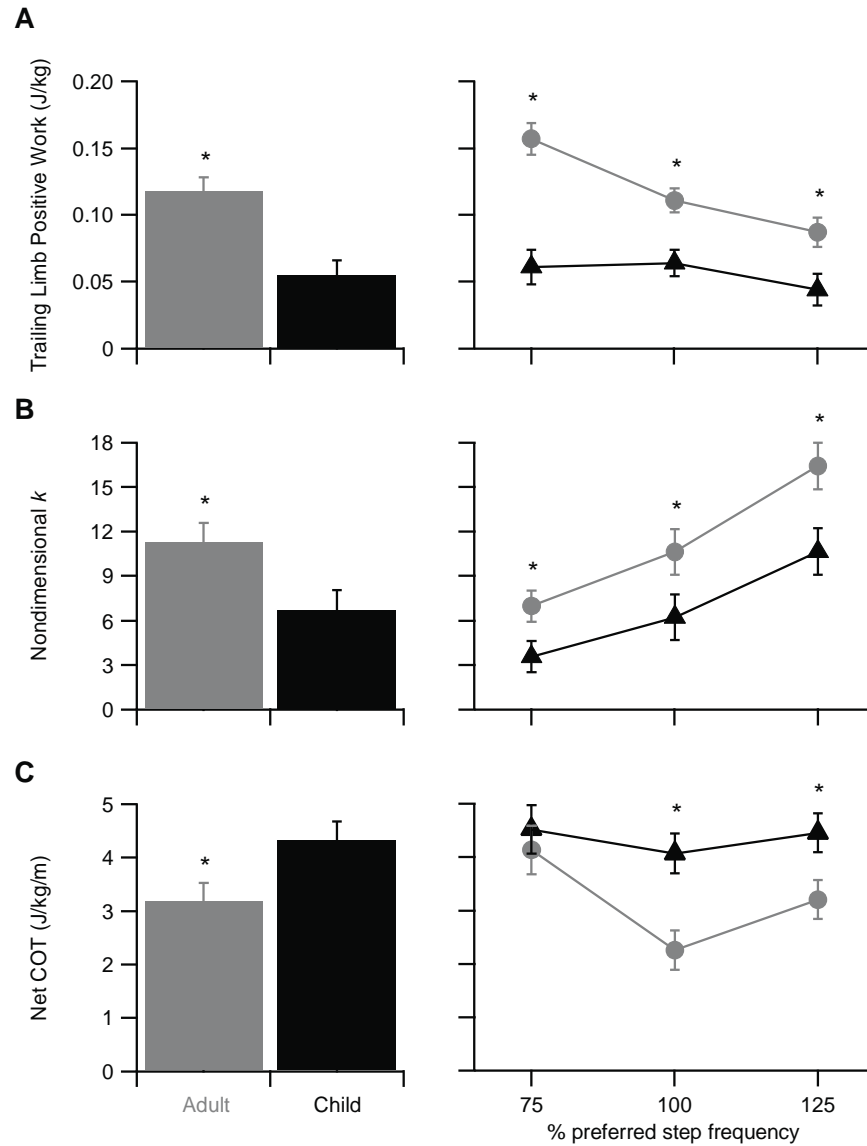
655 **Figure 3.** Representative adult (left, gray) and child (right, black) force and spring length curves during
656 single support phase at preferred step frequency. (A) 2D resultant GRF scaled to bodyweight (bw) and
657 (B) 2D spring length as measured from the COM to centre of pressure, scaled to standing height of COM
658 (l_0) (C) Force-length curves show greater asymmetry and a flatter slope for a child subject. Positive
659 mechanical work at rebound (early single support) reflects recoil of the leg spring just after collision,
660 while negative mechanical work in preload (late single support) reflects compression of the leg spring in
661 preparation for propulsion. The convention here is that the absolute value of the slope equals the
662 normalized spring stiffness, \square .

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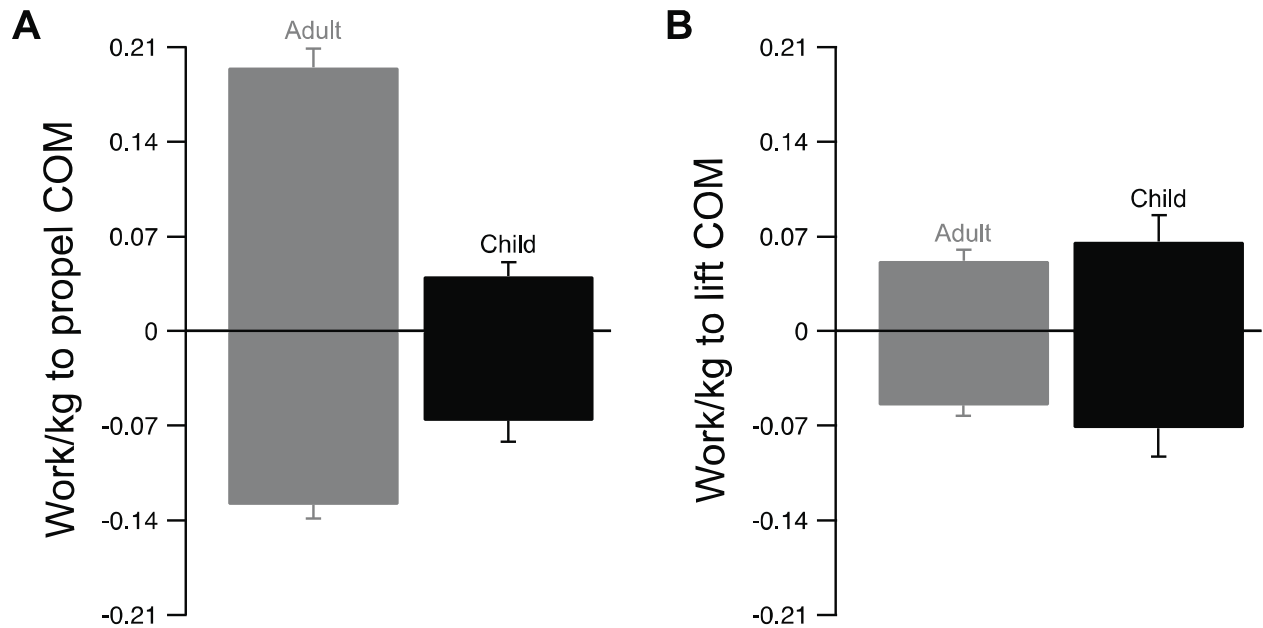
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668 **Figure 4.** (A) The positive work generated by the trailing limb was smaller in children as compared to
669 adults (left; *significant group effect $p=0.039$). In response to walking at relatively slow, preferred, and
670 fast step frequencies, adults (grey) altered the positive work generated by the trailing limb, however,
671 children (black) showed little to no change in trail limb positive work (right; *significant interaction
672 effect $p=0.013$). (B) When compared to adults, children operated their leg stiffness with a lower \hat{k} (left;
673 *significant group effect $p < 0.001$); however, when walking across a range of relatively slow and fast
674 step frequencies, both children and adults modulated \hat{k} in the same way (right; no interaction effect p
675 $=0.41$). (C) Differences in trailing limb positive work and \hat{k} were associated with differences in the net
676 COT required to walk. Overall, children walked with a 36% higher net COT (left; *significant group
677 effect $p=0.03$), yet, both children and adults increased their net COT in response to walking at relatively
678 slow and fast step frequencies, exhibiting a U-shaped trend (right; no interaction effect $p>.05$). Note that
679 for the repeated measures ANCOVAs, average values for trail limb positive work and net COT are
680 adjusted at a Froude number equal to 0.3. For all values, the bars represent the standard error of the mean.
681 Significant differences between children and adults at each step frequency (right) are marked by an *,
682 denoting $p < 0.05$.

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689 **Figure 5.** Partitioning the mean mass-specific work of both limbs reveal that children performed (A)
690 much less work to propel the COM than adults, but (B) slightly higher work to lift the COM. As noted in
691 the textNote that the trailing limb in the child group performed both positive and negative work in double
692 support, while the trail limb in the adult group was only used to generate positive work.

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Table 1: Subject Characteristics

	Children	Adults
Sex (M/F)	4/4	4/4
Age (y)	5.43 ± 0.53	26.38 ± 4.40
Height (m)	1.22 ± 0.04	1.72 ± 0.08
Mass (kg)	24.77 ± 5.60	75 ± 17.40
Spring length (m)	0.70 ± 0.06	1.03 ± 0.06
Leg length (m)	0.62 ± 0.03	0.92 ± 0.05
Preferred speed (m/s)	0.52 ± 0.13	1.13 ± 0.18
Froude number	0.20 ± 0.01	0.40 ± 0.01
Step frequency achieved vs. cued (% difference)	2.47 ± 2.73	0.66 ± 1.30

Values are mean ± SD, Percent difference in step frequency is at the preferred frequency condition

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Table 2. Descriptive Statistics, Pairwise and Planned Comparisons

	Step Frequency Conditions								
	75%			100%			125%		
	Mean	SD	N	Mean	SD	N	Mean	SD	N
W_{trail}^+ (J/kg) ^a									
Child	.06	.01	8	.06	.01	8	.04	.01	7
Adult	.16	.01	8	.11	.01	8	.09	.01	8
Cohen's <i>d</i>	10.00			5.00			3.74		
	<i>p</i><.001			<i>p</i>=.01			<i>p</i>=.03		
\hat{k}									
Child	3.57	3.74	7	6.23	5.32	7	10.66	4.24	7
Adult	6.99	1.29	7	10.64	2.33	7	16.43	4.09	7
Cohen's <i>d</i>	1.22			1.07			1.38		
	<i>p</i>=.035			<i>p</i>=.039			<i>p</i>=.024		
Touchdown angle (°) ^a									
Child	72.88	3.95	8	73.96	2.76	8	77.61	3.07	7
Adult	71.33	3.56	8	75.07	2.48	8	77.72	2.96	8
Groucho/Leg length ^b									
Child	.04	.03	8	.03	.01	8	.05	.02	7
Adult	.04	.01	7	.04	.01	7	.05	.01	7
	<i>p</i> =.911			<i>p</i> =.120			<i>p</i> =.585		

a. Values are evaluated at Froude 0.3 with ANCOVAs when main effect for group is significant; comparisons with □ were followed up with *a priori t*-test when significant. **b.** Independent *t*-test

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Table 3 Metabolic R-ANCOVA results, and ANCOVA pairwise comparisons

	<i>Effects^a</i>	<i>F</i>	<i>p</i>
Net Cost of Transport (J/kg/m)	Step frequency	<i>F</i>=3.805, <i>df</i> = 2,26	0.036*
	Group	<i>F</i>=4.003, <i>df</i> = 1,13	0.034*
	Group x Step frequency	<i>F</i> =2.112, <i>df</i> = 2,26	0.071
	Conditions		
	Children and Adults at 75%	<i>F</i> =0.232, <i>df</i> =1,16	0.319
	Children and Adults at 100%^b	<i>F</i>=7.653, <i>df</i> = 1,16	0.008**
	Children and Adults at 125%	<i>F</i>=3.752, <i>df</i> =1,16	0.038*

a. Mean differences evaluated at Froude 0.3; **b.** Comparison violated assumption of homogeneity of variances, so differences analyzed with a Mann-Whitney U test; **p* < .05, ***p* < .01

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