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1 BIOMECHANICS, ANTHROPOMETRY, WORK PHYSIOLOGY

2 Assessing infant carriage systems: ground reaction force implications for gait of the caregiver

3

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19

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24

25 **Abstract**

26 **Objective:** To assess the acute alterations of anterior infant carriage systems on the ground reaction
27 force experienced during over ground walking.

28 **Background:** Previous research has identified the alterations in posture and gait associated with an
29 increased anterior load (external or internal); however the forces applied to the system due to the altered
30 posture during over ground walking have not been established.

31 **Method:** Thirteen mixed gender participants completed forty-five over ground walking trials at a self-
32 selected pace under three loaded conditions (unloaded, semi-structured carrier 9.9kg and structured
33 carrier 9.9kg). Each trial consisted of a fifteen metre walkway, centred around a piezoelectric force
34 platform sampling at 1200 Hz. Differences were assessed between loaded and unloaded conditions
35 and across carriers using paired samples t-tests and repeated measures ANOVA.

36 **Results:** Additional load increased all ground reaction force parameters; however, the magnitude of
37 force changes was influenced by carrier structure. The structured carrier displayed increased force
38 magnitudes, a reduction in the time to vertical maximum heel contact and an increased duration of the
39 flat foot phase in walking gait.

40 **Conclusion:** Evidence suggest that the acute application of anterior infant carriers alters both kinetic
41 and temporal measures of walking gait. Importantly these changes appear to be governed not solely by
42 the additional mass but also by the structure of the carrier.

43 **Application:** These findings indicate carrier structure should be considered by the wearer and may be
44 used to inform policy in the recommendation of anterior infant carriage systems use by caregivers.

45

46 **Key Words:** Biomechanics, Gait, Posture, Kinetics, Loading, Product design.

47 **Précis:** Use of infant carriers has expanded over the past two decades, however little understanding of
48 the impacts of these on the caregiver exists. Results demonstrated increased forces being applied to
49 the wearer as a result of load and carrier structure. Consideration needs to be given in carrier selection
50 and use.

51

52 **Introduction**

53 The use of ergonomic aids, in the form of woven wraps, to assist in the transportation of infants has
54 been and continues to be commonplace in developing countries throughout the world (Glover, 2012;
55 Wu, Huang, & Wang, 2016). This approach has seen significant increase in developed countries over
56 the past two decades (Frisbee and Hennes, 2000; Glover 2012), resulting in the increased availability
57 of commercial infant carriage devices. This trend can be in part attributed to its promotion by parenting
58 organisations such as the National Childcare Trust (2016), Babywearing International (2015) and the
59 Centre for Babywearing Studies (2016). Proposed benefits include convenience, the promotion of
60 physical development, child mental and physical health, safety, and improved health for the wearer
61 (Natural Life Mom, 2012; Sling Babies, 2011). Whilst some of these claims have been supported in the
62 literature, including convenience (Wu et al., 2016), reduction in crying (Hunziker and Barr, 1986) and
63 an increase in infant mother attachment (Tessier et al., 1998; Gathwala, Singh, & Balhara, 2008), little
64 attention has been directed towards the physical health of the caregiver. Specifically, the short- and
65 long-term implications of carrying an infant on the caregiver's posture, gait and structural health.

66 The task of infant carriage is ostensibly one of load carriage, either anteriorly or posteriorly, while the
67 majority of load carriage work examine the effects of posterior load on posture (Atwells, Birrell, Hooper
68 & Mansfield, 2006; Schiffman, Bense, Hasselquist, Gregorczyk & Piscitelle, 2006), gait (Birrell, Hooper
69 & Haslam, 2007; Birrell & Halsam, 2008; Birrell & Halsam, 2010), ground reaction force (Cavanagh &
70 LaFortune, 1980; Hsiang, Jiang, & McGorry, 1998; Lloyd and Cooke, 2000; Ciacci, Di Michelea, & Mern,
71 2010; Birell et al., 2007), fatigue (Qu and Yeo, 2011) and cardiovascular response (Fallowfield, Blacker,
72 Willems, Davey, & Layden, 2012). Application of many of these findings are limited in reference to
73 anterior load carriage, given the significant differences reported by Fiolkowski, Horodyski, Bishop,
74 Williams, and Stylianou (2006) in gait kinematics between anterior and posterior loads. However,
75 findings associated with cardiovascular response, namely the increased energy cost associated with
76 an additional load, as measured by oxygen consumption and heart rate (Fallowfield et al., 2012), and
77 increase in forces experienced proportionate to the load applied (Birrell et al., 2007) are more readily
78 transferable. Consequently, the use of an anterior infant carriage system could have cardiorespiratory
79 adaptations resulting in enhanced health and reduced disease risk, supporting the claims of parenting

80 groups associated with the wearer's health (Natural Life Mom, 2012; Sling Babies, 2011). However,
81 focus on anterior load carriage has been sparse in the academic literature with reference to posture
82 and gait parameters (Hsiang et al., 1998; Birrell and Haslam, 2008; Junqueira, Amaral, Lutaka, &
83 Duarte, 2015; Fiolkowski et al., 2006; Perry, et al., 2010; Graham, Smallman, Miller, & Stevenson,
84 2014). Findings indicate that anterior load carriage, using a front pack equivalent to 10 and 15% of
85 participant mass, caused a reduced hip flexion and extension, compared to unloaded walking
86 (Fiolkowski et al., 2006). While application of a fixed 4.4kg load (divers belt) identified a significant
87 decrease in vertical ground reaction force at maximum vertical thrust at push off (Birrell and Haslam,
88 2008), no other vertical force measures were significantly altered by the fixed anterior load. This may
89 have been a result of the alteration in the centre of gravity caused by the anterior mass, reducing the
90 impulse needed to accommodate the load at push off (Hsiang et al., 1998). Furthermore, Junqueira et
91 al. (2015) identified significant alterations in trunk orientation when participants carried live infants and
92 infant mannequins in their arms. These were characterised by increased trunk inclination, lumbar
93 lordosis and thoracic kyphosis during standing posture and walking (Junqueira et al., 2015).

94 Associated literature addressing impact of increased anterior load can be found in analysis of gait during
95 pregnancy; Junqueira et al. (2015) highlighted commonalities in qualitative movement patterns of
96 pregnant gait and postpartum infant carriage gait. Furthermore, significant reduction in walking velocity
97 during pregnancy (McCrary, Chambers, Daftary, & Redfern, 2011) and postpartum infant carriage
98 (Junqueira et al., 2015) has been established. These alterations are suggested to be compensatory to
99 mitigate the increased instability of the caregiver/child system caused by changes in the position of the
100 centre of gravity (Branco, Santos-Rocha, & Vieira, 2014). Consequently, these similarities may indicate
101 that mothers are well positioned to transfer from in vivo carriage to postpartum carriage; however, non-
102 maternal caregivers will have had no such adaptations. Furthermore, the TICKS guidelines developed
103 by the Consortium of UK Sling Manufacturers (National Childbirth Trust, 2016) state that the child should
104 be positioned high on the chest close enough for the carer to kiss the child on the forehead. In
105 comparison to previous research, where loads were carried in a much lower position, the centre of
106 gravity will be raised and therefore the alterations in gait characteristics further exaggerate.

107 Considering the alterations in walking posture and kinematics (Fiolkowski et al., 2006; Junqueira et al.,
108 2015) understanding the loading of the body is important, as joints and muscles will be loaded outside
109 of the general motor pattern, exposing the wearer to increased prospects of injury (Bonci, 1999). This

110 could be magnified by a lack of pregnancy adaptations in non-maternal caregivers, therefore the use of
 111 anterior infant carriage systems could have health implications for caregivers. In light of the limited work
 112 toward understanding the impact of anterior load, and specifically, that no research has yet established
 113 the impact of anterior infant carriage on the caregiver, the aim of the current research was to ascertain
 114 the changes in ground reaction forces experienced when carrying an anterior load on the chest using
 115 an infant carrier. Moreover, it aims to determine acute alterations in temporal and kinetic parameters of
 116 the foot ground interaction experienced by the caregiver during walking, and if this is affected by specific
 117 carrier structure.

118

119

120

121 **Method**

122 Participant Recruitment

123 Thirteen injury free participants (female = 7; male = 6; mean age = 29.3 ± 8.65 years) volunteered to
 124 take part in this study. Participant demographics are presented in Table 1. This research complied
 125 with the tenets of the Declaration of Helsinki and was approved by the Institutional review board at
 126 Canterbury Christ Church University. Informed consent was obtained from each participant. Inclusion
 127 criteria required all participants to be free from injury at time of data collection and have had no back,
 128 lower limb or shoulder injuries in the previous 12 months. No participants had given birth in the
 129 previous 12 months.

130 Table 1. Participant Demographics

Gender	No. of Participants	Descriptive Statistic	Age (years)	Height (cm)	Mass (kg)	BMI (kg/m ²)
Female	7	Min	23	162.30	57.60	19.80
		Max	48	174.60	90.70	31.24
		Mean	31.71	168.83	69.06	24.23
		SD	10.84	5.20	11.74	3.94
Male	6	Min	23	179	65.30	19.67
		Max	35	190.20	95	27.31
		Mean	26.5	184.37	80	23.49
		SD	4.59	4.29	11.67	2.90

131

132 No previous research was deemed acceptable for an a priori sample size estimation, therefore post-
133 hoc power analyses were conducted for the repeated measures T-test and repeated measures
134 ANOVA. Using G*Power (v. 3.19.2) with an alpha of 0.05 and a large effect size (0.8), demonstrated
135 power was 0.75 and 0.9 for the t-tests and ANOVA respectively.

136 Experimental Conditions

137 Participants completed fifteen barefoot trials at a self-selected walking pace over a 15 m distance in
138 each condition (unloaded $1.54 \pm 0.03\text{m}\cdot\text{s}^{-1}$; and 2 anteriorly loaded conditions, structured [SC] $1.54 \pm$
139 $0.03\text{m}\cdot\text{s}^{-1}$; semi-structured [SSC] $1.52 \pm 0.03\text{m}\cdot\text{s}^{-1}$), making contact with their right foot on a force
140 platform. Barefoot conditions were used to ensure that differences between shoe construction and
141 condition between participants did not affect force measures, as these have been demonstrated to
142 influence force attenuation and foot and ankle kinematics during gait (Novacheck, 1998).

143 Table 2. Specifications of Infant Carriage Systems

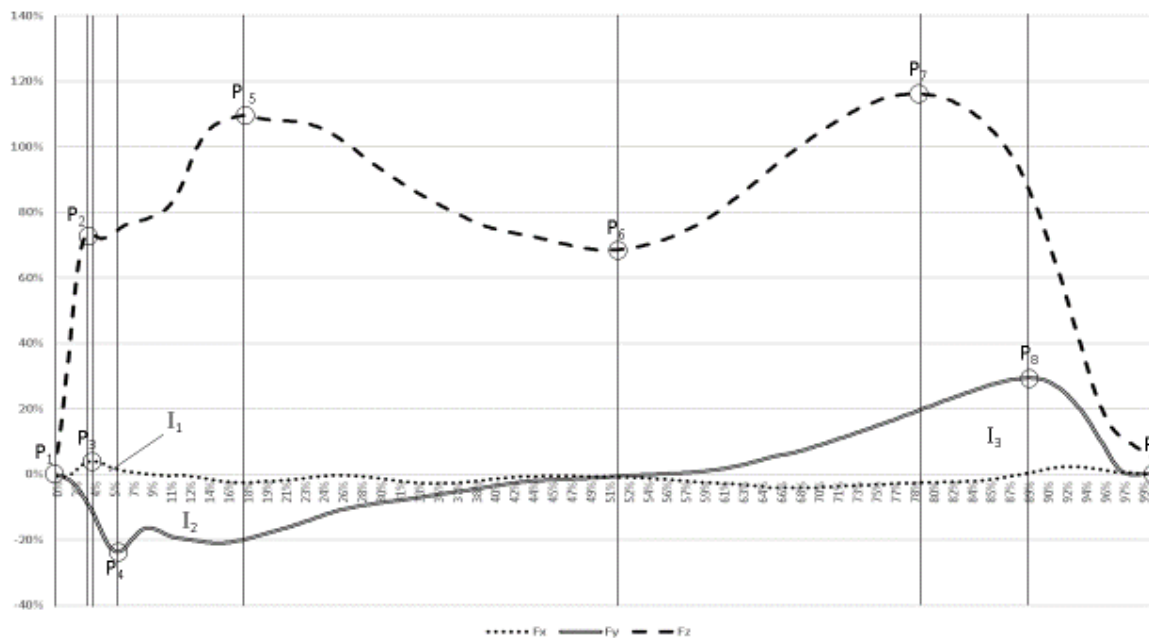
Infant Carrier	Semi Structured	Structured
Picture		
Weight (g)	576	997
Material	100% cotton	Main material: 60% cotton, 40% polyester Lining: 100% cotton Waist belt: 100% polyester Mesh: 100% polyester Cover for leg position zip: 100% cotton
Product Features	A comfortable and supportive baby carrier that allows you to carry on your front, hip or back. Allows the carrier to grow with your child to fit any size of baby or toddler from newborn up to 4 years old. Baby is securely supported in the best position for healthy hip development. Wide shoulder straps to evenly spread the weight around your	Ergonomic baby carrier with wide seat area Extra-padded shoulder straps Good stability in the waist belt Perfect for a newborn – no infant insert needed. Front-facing carrying option From newborn to 3 years Acknowledged as a hip-healthy baby carrier by International Hip Dysplasia Institute

	body and provide a custom fit for each user. Padded waist for extra comfort.	
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144

145 The unloaded condition was completed barefoot, wearing minimal clothing, defined as tight fitting top
 146 and sports shorts and no carrier. The loaded conditions consisted of the wearing of 2 different
 147 anteriorly loaded infant carriers, one semi-structured and one structured, specifications for which can
 148 be found in Table 2. Both carriers were loaded with a purpose-made mannequin with the equivalent
 149 mass of a 12 month old on the 50th percentile on the NHS growth charts (9.9 kg) (NHS, 2017). A
 150 mannequin was used as it has previously been reported (Junqueira et al., 2015) that the carriage of a
 151 mannequin results in similar alterations in walking kinematics to carrying one's own infant when
 152 compared to unloaded walking. Participants were instructed to allow their arms to swing naturally
 153 during walking trials, rather than holding on to the mannequin and carrier. The order in which the
 154 participants completed each condition was randomised using an online research randomizer
 155 (Urbaniak and Plous, 2013).

156



157

158 *Figure 1. Annotated typical ground reaction force trace. P1 – Initial Contact, P2 – Impact Force Peak, P3 – Medial Peak*
 159 *Force, P4 – Max Posterior Braking, P5 – Max Heel Contact, P6 – Midstance, P7 – Max Vertical Thrust, P8 – Max Anterior*
 160 *Propulsive, P9 – Toe Off, I1 – Medial Impulse, I2 – Braking Impulse, I3 – Propulsive Impulse.*

161

162 Instrumentation

163 Ground reaction force (GRF) data were sampled at 1200 Hz via a 900 x 600 mm Kistler force platform
164 (Model 9287BA, Kistler Instruments Ltd) using Bioware software (v5.3.0.7, Kistler Instruments Ltd).
165 Timing lights (in house, Canterbury Christ Church University) recorded the time taken to complete the
166 central 5 m of the over ground walking trials, centred over the force platform, to allow calculation of
167 walking velocity.

168

169 Data Analysis

170 Data files containing GRF components for over ground walking were filtered in Bioware (v5.3.0.7,
171 Kistler Instruments Ltd) using a dual-pass Butterworth low-pass filter with a cut-off frequency of 50 Hz
172 (McCrorry et al., 2011; McCrorry, Chambers, Daftary, & Redfern, 2013). A Fast Fourier Transformation
173 of 13 randomly selected trials revealed data to be below 45 Hz. GRF data files were exported from
174 Bioware to Excel, where a purpose-written analysis template extracted key kinetic and temporal
175 components for further analysis. Peak vertical and anteroposterior force and impulse were calculated
176 as key events in the loading of the gait cycle and have been demonstrated to be important responders
177 to assess force during general load carriage (Birrell, 2007). Rates of force loading and unloading were
178 included to assess acceleration changes to the caregiver and carriage system, beyond that of
179 maximal amplitudes, as indication of increased injury risk (Greenhalgh, Sinclair, Protheroe and
180 Chockalingham, 2011). The mediolateral assessment was included, as despite the small magnitude
181 of these during walking gait, research suggests mediolateral stability is important in similar anterior
182 load carriage tasks (Branco et al., 2013; Branco et al., 2014; LyMBERY and GILLEARD, 2005) and
183 therefore was deemed important for inclusion. Calculations for all variables are outlined in Tables 3
184 and 4, figure 2 and with reference to figure 1, except for rate from peak medial force to max lateral
185 force was calculated using maximums prior to midstance (MS). The fastest and the slowest walking
186 trials were removed from analysis, leaving 13 trials per condition per participant, and GRF data were
187 normalised to body weight (BW), and temporal measures were normalised to contact time.

188 Statistical Analysis

189 Data were checked for normality using a Shapiro-Wilk test. Pairwise assessment of loaded
190 (combination of the two loaded conditions) versus unloaded, and between carrier conditions were
191 conducted to ascertain if carrier structure was associated with any significant differences in GRF.
192 Pairwise comparison between loaded and unloaded were calculated using either a paired t-test or a
193 Wilcoxon matched-pairs test, from which a Cohen's d effect size was calculated and interpreted as
194 small (0.2), medium (0.5) and large (0.8). Cross carrier assessment employed either a repeated
195 measures ANOVA across carrier condition, where significant differences were ascertained through
196 pairwise comparison using a Bonferroni Post Hoc analysis, or a Friedman Test. Effect size for the
197 repeated measures ANOVA was calculated using partial eta-squared and interpreted as small (0.01),
198 medium (0.09) and large (0.25). The alpha level for all tests was set to 0.05 and all tests were carried
199 out using SPSS (version 22, IBM, NY).

200

201 **Results**

202 *Unloaded vs. Loaded*

203 *Kinetic Analysis*

204 Pairwise comparisons between unloaded and loaded conditions (Table 3) identified significant
205 increases in impact force peak (IFP: $t(10) = -3.243$, $p = 0.009$, $d = 0.98$, Un: 0.745 ± 0.034 , Loaded:
206 0.869 ± 0.054), maximum heel contact (MHC: $t(12) = -11.307$, $p = 0.000$, $d = 3.14$, Un: 1.206 ± 0.032 ,
207 Loaded: 1.375 ± 0.034), midstance (MS: $t(12) = -10.752$, $p = 0.000$, $d = 2.98$, Un: 0.663 ± 0.019 ,
208 Loaded: 0.772 ± 0.019) and maximum vertical thrust (vertical force component propulsive peak;
209 MaxT: $t(12) = 14.714$, $p = 0.000$, $d = 4.08$, Un: 1.149 ± 0.012 , Loaded: 1.275 ± 0.017) under loaded
210 conditions. Rate of vertical force loading at heel contact (VLR) and rate of force unloading (VLOR) at
211 end of stance also significantly increased when loaded ($t(12) = -3.890$, $p = 0.002$, $d = 1.08$; $t(12) =$
212 6.283 , $p = 0.000$, $d = 1.74$, respectively).

213 Similarly, maximum posterior braking force (MPB: $t(12) = 4.566$, $p = 0.001$, $d = 1.27$) and maximum
214 anterior propulsive force (MAP: $t(12) = -6.734$, $p = 0.000$, $d = 1.87$) significantly increased under
215 loaded conditions. Loaded walking also resulted in significant increases in braking impulse (BI) and
216 propulsive impulse (PI: $t(12) = 8.921$, $p = 0.000$, $d = 2.47$; $t(12) = -7.852$, $p = 0.000$, $d = 2.18$,

217 respectively), however rate of braking force (BLR) application was not significantly altered ($z = -1.645$,
218 $p = 0.101$). Furthermore, load significantly increased the medial peak force (MPF: $t(12) = -2.386$, $p =$
219 0.034 , $d = 0.66$) and medial loading rate (MLR: $z = -2.481$, $p = 0.013$, $d = 0.69$), however medial
220 impulse (MI) did not alter significantly.

221 In the transition from MHC to MaxT (the flat foot phase of stance), significant increases were evident
222 in the magnitude of force changes between MHC and MS ($t(12) = -2.812$, $p = 0.016$, $d = 0.78$) and
223 MHC and MaxT ($t(12) = -3.156$, $p = 0.008$, $d = 0.88$) under loaded conditions. However these
224 changes were not sufficient to significantly alter the load off rate between MHC and MS and the rate
225 of force application from MS to MaxT ($p > 0.05$) or the force magnitude change between MS and MaxT.

226 Table 3. Kinetic Analysis

		Unloaded	Loaded (combined)	Semi-Structured (SSC)	Structured (SC)
<i>Peak Force (BW)</i>					
P_4	F _y - Max Posterior Braking * $\alpha\beta\gamma$	-0.250 ± 0.008	-0.272 ± 0.008	-0.269 ± 0.009	-0.275 ± 0.008
P_8	F _y - Max Anterior Propulsive * $\alpha\beta\gamma$	0.271 ± 0.007	0.307 ± 0.008	0.304 ± 0.008	0.309 ± 0.009
P_3	F _x - Medial Peak Force * $\alpha\gamma$	0.068 ± 0.007	0.075 ± 0.008	0.074 ± 0.008	0.076 ± 0.008
<i>Impulses (BW·s)</i>					
I_1	F _x - Medial Impulse	0.0020 ± 0.0002	0.0022 ± 0.0002	0.0022 ± 0.0002	0.0022 ± 0.0002
I_2	F _y - Braking Impulse * $\alpha\beta\gamma$	-0.0371 ± 0.0011	-0.0420 ± 0.0013	-0.0418 ± 0.0015	-0.0422 ± 0.0011
I_3	F _y - Propulsive Impulse * $\alpha\beta\gamma$	0.0358 ± 0.0013	0.0406 ± 0.0013	0.0404 ± 0.0013	0.0409 ± 0.0013
<i>Loading Rates (BW·s⁻¹)</i>					
$\frac{Fz_{P_2} - Fz_{P_1}}{t_{P_2} - t_{P_1}}$	F _z - Impact Loading Rate * $\alpha\beta\gamma$	51.285 ± 3.430	56.256 ± 3.680	55.678 ± 3.687	56.841 ± 3.746
$\frac{Fz_{P_9} - Fz_{P_7}}{t_{P_9} - t_{P_7}}$	F _z - Load Off Rate * $\alpha\beta\gamma$	-16.615 ± 0.588	-18.725 ± 0.718	-18.413 ± 0.706	-19.038 ± 0.763
$\frac{Fx_{P_3} - Fx_{P_1}}{t_{P_3} - t_{P_1}}$	F _x - Medial Impact Loading Rate * $\alpha\gamma$	3.184 ± 0.458	3.681 ± 0.577	3.632 ± 0.547	3.730 ± 0.611
-	F _x - Max Med. to Max Lat. Rate $\alpha\beta$	1.257 ± 0.235	1.428 ± 0.216	1.450 ± 0.207	1.406 ± 0.232
$\frac{Fz_{P_6} - Fz_{P_5}}{t_{P_6} - t_{P_5}}$	F _z - MHC to MS Load Off Rate $\alpha\gamma\delta$	-3.436 ± 0.298	-3.654 ± 0.299	-3.525 ± 0.304	-3.784 ± 0.307
$\frac{Fz_{P_7} - Fz_{P_6}}{t_{P_7} - t_{P_6}}$	F _z - MS to MaxT Load Rate	2.658 ± 0.164	2.734 ± 0.145	2.716 ± 0.156	2.751 ± 0.141
$\frac{Fy_{P_4} - Fy_{P_1}}{t_{P_4} - t_{P_1}}$	F _y - Braking Force Rate	-3.681 ± 0.379	-4.064 ± 0.493	-3.950 ± 0.477	-4.178 ± 0.516
<i>Delta Changes (BW)</i>					
$P_6 - P_5$	F _z - MHC - MS Difference * $\alpha\gamma$	0.543 ± 0.050	0.603 ± 0.050	0.585 ± 0.051	0.622 ± 0.050
$P_5 - P_7$	F _z - MHC - MaxT Difference * $\alpha\gamma$	0.057 ± 0.034	0.100 ± 0.038	0.089 ± 0.037	0.112 ± 0.039
$P_7 - P_6$	F _z - MS - MaxT Difference	0.486 ± 0.025	0.503 ± 0.023	0.496 ± 0.025	0.510 ± 0.022

227 *Denotes significant difference between loaded and unloaded condition; α denotes a significant finding from Repeated Measures ANOVA; β denotes significant pairwise comparison between Unloaded and Semi-
 228 Structured; γ denotes significant pairwise comparison between Unloaded and Structured; δ denotes significant pairwise comparison between Semi-Structured and Structured.

229 *Temporal Analysis*

230 A significant ($t(12) = -2.260, p = 0.043, d = -0.63$) increase in time between the MHC and MaxT in the
 231 loaded condition ($57.81 \pm 0.57\%$) in comparison to unloaded ($56.91 \pm 0.47\%$), indicated that the load
 232 increased the time during which participant's full foot was in contact with the force platform. No
 233 significant difference between loaded/unloaded conditions were observed for other temporal
 234 measures (Table 4).

235 Table 4. Temporal Analysis

Calculation		Unloaded	Loaded (combined)	Semi-Structured	Structured
$t_{P_9} - t_{P_1}$	Contact Time (s)	0.604 ± 0.009	0.607 ± 0.008	0.608 ± 0.008	0.606 ± 0.009
$\frac{t_{P_2} - t_{P_1}}{t_{P_9} - t_{P_1}}$	Time to Impact Peak (% CT)	3.95 ± 0.55	4.26 ± 0.91	4.49 ± 1.08	4.02 ± 0.80
$\frac{t_{P_5} - t_{P_1}}{t_{P_9} - t_{P_1}}$	Time to Max Heel Contact (% CT) ^α	21.29 ± 0.44	20.65 ± 0.46	20.87 ± 0.53	20.43 ± 0.42
$\frac{t_{P_6} - t_{P_1}}{t_{P_9} - t_{P_1}}$	Time to Midstance (% CT)	47.34 ± 0.61	47.87 ± 0.71	48.11 ± 0.74	47.64 ± 0.71
$\frac{t_{P_6} - t_{P_5}}{t_{P_9} - t_{P_1}}$	Time to MS from MHC (% CT)	26.05 ± 0.63	27.22 ± 0.72	27.23 ± 0.72	27.21 ± 0.72
$\frac{t_{P_7} - t_{P_1}}{t_{P_9} - t_{P_1}}$	Time to MaxT (% CT)	78.20 ± 0.43	78.46 ± 0.45	78.38 ± 0.48	78.53 ± 0.43
$\frac{t_{P_7} - t_{P_5}}{t_{P_9} - t_{P_1}}$	Time to MT from MHC (% CT) ^{*αγ}	56.91 ± 0.47	57.81 ± 0.57	57.51 ± 0.61	58.11 ± 0.56
$\frac{t_{P_7} - t_{P_6}}{t_{P_9} - t_{P_1}}$	Time to MaxT from MS (% CT)	30.62 ± 0.57	30.55 ± 0.55	30.25 ± 0.55	30.86 ± 0.61
$\frac{t_{P_9} - t_{P_7}}{t_{P_9} - t_{P_1}}$	Time from MaxT to Toe off (% CT)	21.80 ± 0.43	21.54 ± 0.45	21.62 ± 0.48	21.47 ± 0.43
-	Time Max medial Force to Max Lateral Force (%CT)	20.64 ± 1.92	19.45 ± 1.60	19.35 ± 1.77	19.56 ± 1.55
-	Velocity Final ($m \cdot s^{-1}$)	1.538 ± 0.031	1.532 ± 0.031	1.519 ± 0.030	1.545 ± 0.032

236 *Denotes significant difference between loaded and unloaded condition; ^α denotes a significant finding from Repeated Measures ANOVA; ^β
 237 denotes significant pairwise comparison between Unloaded and Semi-Structured; ^γ denotes significant pairwise comparison between
 238 Unloaded and Structured; ^δ denotes significant pairwise comparison between Semi-Structured and Structured.

239

240 *Analysis by Carrier Type*

241 *Kinetic Analysis*

242 Significant findings in all vertical force measures from the paired samples t-tests were duplicated in
243 the overall effect of the repeated measures ANOVA when load was separated by carrier type. Post
244 hoc pairwise comparisons identified the magnitude of the MHC ($F(2,24) = 96.589, p < 0.001, \eta^2 =$
245 0.89) was significantly higher under the SC condition (1.390 ± 0.034 Bw), with MHC diminishing
246 through the SSC condition (1.361 ± 0.035 Bw) to the unloaded condition (1.206 ± 0.032 Bw). All other
247 vertical force measures exhibited no difference between SSC and SC, this held true for the peak
248 forces and impulses in anterior posterior forces. Medial impulse ($F(1.359,16.309) = 1.9, p = 0.171, \eta^2 =$
249 0.137) and MPF ($F(1.399,16.794) = 4.91, p = 0.031, \eta^2 = 0.29$) also echoed the paired samples
250 analysis, however Bonferroni post hoc pairwise analysis of the MPF was too conservative to identify
251 the source of the significant difference. Further investigation using a lowest significant difference
252 (equivalent to no adjustments), identified the source of the difference, with the SC condition ($0.076 \pm$
253 0.008 Bw) MFP magnitude being significantly higher than the unloaded condition (0.068 ± 0.007 Bw).
254 Resultantly, the same pattern of significance was identified in the MLR ($\chi^2(2) = 6.615, p = 0.037$) and
255 the Wilcoxon post hoc identified significance between SC and Unloaded ($z = -2.481, p = 0.013, d =$
256 0.69).

257 The transition from MHC to MaxT highlighted further impacts of the SC carrier condition. The
258 magnitude change between MHC and MS ($F(2,24) = 7.267, p = 0.003, \eta^2 = 0.38$) was significantly
259 impacted by load condition, post hoc testing highlighted a significantly greater drop from MHC to MS
260 (SC: 0.622 ± 0.050 , Unloaded: 0.543 ± 0.050). This led to a significant increase in the load off rate
261 from MHC to MS ($F(2,24) = 3.707, p = 0.040, \eta^2 = 0.236$). Interestingly, the SC condition ($-3.784 \pm$
262 0.307 Bw.s⁻¹) was significantly faster than the unloaded (-3.436 ± 0.298 Bw.s⁻¹), but not the SSC ($-$
263 3.525 ± 0.304 Bw.s⁻¹, $p = 0.051$) conditions, although this findings required the use of the least
264 significant differences approach due to the conservative nature of the Bonferroni post hoc previously
265 identified.

266 The magnitude of change between MHC and MaxT showed further significance alterations due to
267 increased anterior load ($f(2,24) = 8.201, p = 0.002, \eta^2 = 0.406$): again the SC condition ($0.112 \pm$
268 0.039 Bw) was significantly higher than unloaded (0.057 ± 0.034 Bw).

269

270 *Temporal Analysis*

271 A significant difference between conditions for the time between MHC and MaxT as a percentage of
272 contact time ($F(2, 24) = 5.152, p = 0.014, \eta^2 = 0.30$) was apparent. Post hoc pairwise comparisons
273 identified a significant difference between unloaded (56.91 ± 0.47 %CT) and the SC (58.11 ± 0.56
274 %CT) condition; the SSC condition exhibited no significant difference (57.51 ± 0.61 %CT). While the
275 ANOVA for time from heel contact to MHC indicated significant differences ($F(2,24)= 3.475, p =$
276 $0.047, \eta^2 = 0.23$), Bonferroni post hoc tests were too conservative to identify the specific source of the
277 difference reported. Further examination using a lowest significant difference (equivalent to no
278 adjustments) post hoc assessment, identified the significant difference between UN (21.29 ± 0.44
279 %CT) and SC (20.43 ± 0.42 %CT) conditions ($p = 0.040$). Examination of the data indicates that both
280 the loaded conditions (SSC 20.87 ± 0.53 %CT; SC 20.43 ± 0.42 %CT) were characterised by a faster
281 move to MHC than the UN condition (21.29 ± 0.44 %CT), although not enough to significantly alter
282 loading rate at impact.

283

284 **Discussion**

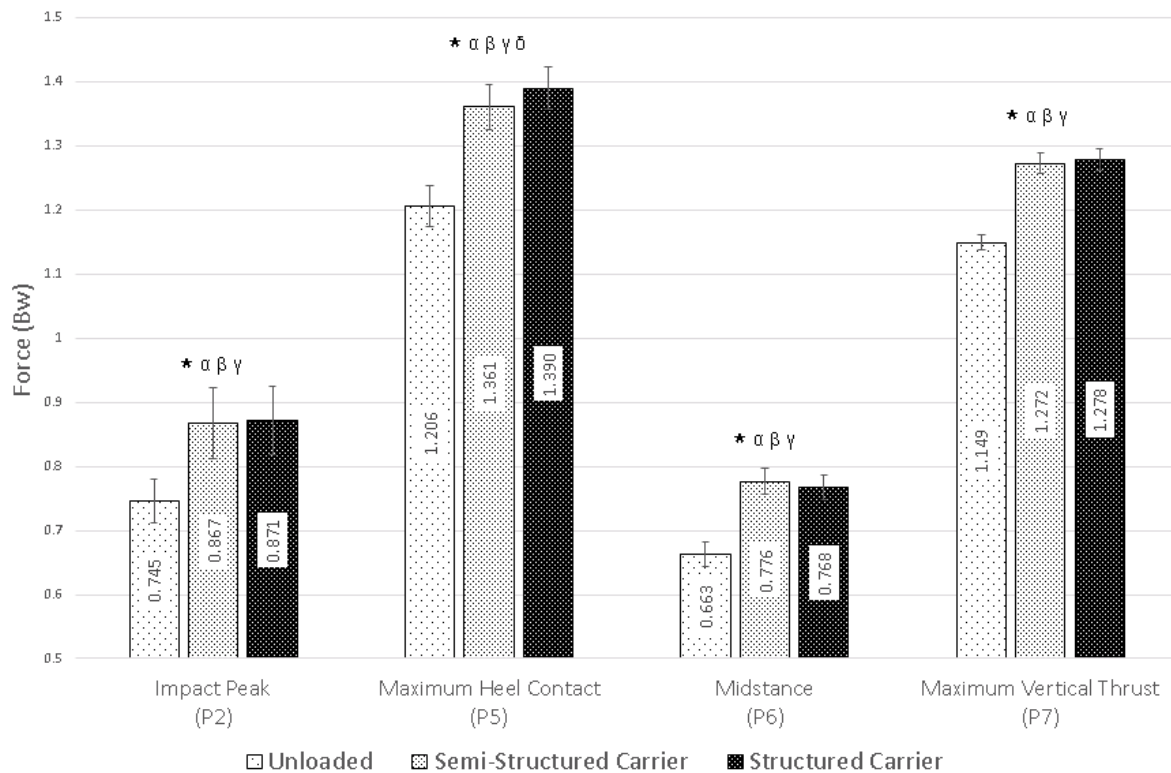
285 Previous studies have identified a decrease in walking velocity in response to the addition of an
286 external anterior load (Junqueira et al., 2015) and internal anterior mass (McCrorry et al., 2011). Our
287 findings do not support this. While this contradiction was unexpected, previous work used mothers
288 only and had the infant (or mannequin) supported in the arms, conversely the focus of this work was
289 to investigate the caregiver (non-gender specific) and investigated the effect of an ergonomic aid
290 (carrier) to assist in the carriage task.

291 Statistically significant increases in peak vertical force parameters (IP, MHC, MS, MaxT) were
292 demonstrated in both loaded conditions when compared with unloaded (figure 2). These increases
293 were in direct opposition to Birrell and Haslam (2008) who found a significant reduction in MaxT in
294 response to the load, with all other measures demonstrating no change. This contradiction is likely
295 due to the increased load used in the current study (9.9kg vs. 4.4 kg Birrell & Haslam, 2008), and
296 could be influenced by the raised position of the anterior mass. Interestingly, the significant increase

327 in MaxT, combined with the significant increase in propulsive impulse ($t(12) = 8.921, p = 0.000, d =$
328 2.47), also contradicted the propositions of Hsiang et al. (1998), who suggested a decreased impulse
329 was required at push off with an anterior load. These contradictions could have resulted from the
330 nature of the task, as previous research had addressed a load (Hsiang et al., 1998; Birrell and
331 Haslam, 2008), where the current research was that of carrying an infant. While this proposition
332 cannot be proved due to the psychological influences being outside of the scope of this paper,
333 participants in the current study may have used an external focus of attention and as such altered the
334 gait accordingly. As it has been shown that this approach can effect motor patterns (Wulf, Weigelt,
335 Poulter, & McNevin, 2003) and Junquera et al. (2015) identified that mannequin carriage and infant
336 carriage demonstrate similar variations from unloaded walking.

337 Increases early in foot contact, specifically the MHC, displayed variation beyond that of the load
338 alone. The use of a structured carrier resulted in a significantly higher force (SC: 1.390 ± 0.034 Bw)
339 being experienced by the caregiver compared to the SSC and unloaded conditions (SSC: $1.361 \pm$
340 0.035 Bw, Unloaded: 1.206 ± 0.032 Bw), this was deemed to have a large effect ($\eta^2 = 0.89$). This may
341 have been influenced by the significant reduction in time between heel contact and MHC ($F(2,24)$
342 $= 3.475, p = 0.047, \eta^2 = 0.225$). Post hoc assessment identified significance values of 0.070 between
343 SC and SSC and 0.040 between SC and unloaded, indicating that the use of the SC is characterised
344 by a quicker transfer from heel contact to MHC through increased acceleration, and therefore
345 resulting in the increased force measured at MHC. When considered alongside the findings of
346 Fiolkowski et al. (2006) and Junqueira et al. (2015), both of whom applied loads of similar magnitude
347 to the current study (15% and 10kg, respectively), and Bonci (1999), this indicates that the caregiver
348 is being exposing to increased stresses and possibly enhanced risk of injury due to greater magnitude
349 of all vertical forces, reduced time to maximum load (MHC) and the alterations in posture previously
350 described.

351



322

323 *Figure 2. Vertical force parameter changes due to load and carrier type. *Denotes significant difference between loaded and*
 324 *unloaded condition; ^a denotes a significant finding from Repeated Measures ANOVA, indicating difference between the 3 conditions; ^b*
 325 *denotes significant pairwise comparison between Unloaded and Semi-Structured; ^γ denotes significant pairwise comparison between*
 326 *Unloaded and Structured; ^δ denotes significant pairwise comparison between Semi-Structured and Structured.*

327 The significantly higher MHC peak force resulted in further significant alterations in the loading
 328 patterns during walking. The magnitude of the reduction from MHC to MS was significantly larger
 329 under loaded conditions ($t(12) = -2.812, p = 0.016, d = 0.78$). Analysis by carrier identified statistical
 330 significance between SC and Unloaded ($p = 0.022$). This linked to further significant differences
 331 between loaded and unloaded ($F(2,24) = 3.707, p = 0.040, \eta^2 = 0.24$), with the rate of force unloading
 332 from MHC to MS being significantly faster under SC conditions when compared to the unloaded
 333 condition (SSC: $p = 0.051$), placing the caregiver under greater extremes of force. As a result, this
 334 may increase the likelihood of injury, if the carrier was employed for prolonged use, or could positively
 335 affect the caregiver through the overload principle attributed to resistance training (Winett & Carpinelli,
 336 2001). Further research addressing prolonged use is needed to ascertain the veracity of these
 337 propositions.

338 The combination of the decrease in time from heel contact to MHC and maintenance of the contact
 339 time ($t(12) = -0.558, p = 0.587$) resulted in a significant increase in duration, as a percentage of
 340 contact time, between MHC to MaxT, with participants spending significantly longer in this transition

341 when wearing the infant carriage systems. When carriage systems were separated ($F(2,24) = 5.152$, p
342 $= 0.014$, $\eta^2 = 0.30$), the foundation of this increase was the SC, exhibiting an increase of 1.2%. This
343 increase in time indicates a longer period of knee flexion during stance, requiring greater muscular
344 effort, especially given the increased forces experienced under loaded conditions. Furthermore,
345 increasing the time in the flat foot phase of stance may be indicative of participants attempting to
346 stabilise the system. It has been reported through kinematic analysis that stability of the body is the
347 primary focus of the pregnant woman during gait (Branco, Santos-Rocha, Aguiar, Vieira, & Veloso,
348 2013; Branco et al. 2014; Lymbery & Gilleard, 2005), and thus it could be expected that the same
349 would be true of postpartum mothers and other caregivers when carrying their infant in a carrier
350 system. While little previous research has identified kinetic changes (Lymbery & Gilleard, 2005), what
351 has been identified supports the importance of stability. Although the temporal findings do not directly
352 speak to previous research addressing stability, as emphasis has been placed on the importance of
353 stability in the medio-lateral direction (Branco et al., 2013, Branco et al. 2014, Lymbery & Gilleard,
354 2005), they do indicate that an acute adaptation has been employed by participants resulting in a
355 longer duration of flat foot contact a greater stability.

356 Analysis of the medio-lateral parameters from both the pair-wise analyses and repeated measures
357 ANOVA displayed mixed findings. Alterations in medio-lateral parameters were inconsistent; the
358 medial impulse demonstrated no significant alteration due to load or carrier type, where rate of force
359 transfer from medial to lateral peak, MPF, and MLR all displayed significant alterations in response to
360 load. The rate of force transfer from medial to lateral during stance, although significant overall ($\chi^2(2)$
361 $= 6.000$, $p = 0.050$), did not clearly demonstrate the specific source, as post hoc analysis, using alpha
362 level correction, could not ascertain the specific source of the significance. The MPF and
363 consequently MLR increases ($t(12) = -2.386$, $p = 0.034$, $d = 0.66$; $Z = -2.481$, $p = 0.013$, $d = 0.69$,
364 respectively) were solely a function of the SC condition. These findings, combined with those from the
365 temporal and vertical force analysis, indicate that carrier structure in addition to load has an influence
366 on the magnitude of forces experienced by the wearer. Careful consideration is therefore required
367 when selecting an anterior infant carrier.

368

369

370 **Conclusion**

371 Results indicate that the use of an infant carrier caused a significant increase in the magnitudes of the
372 forces experienced during walking and altered the temporal characteristics of caregiver gait.

373 The significant increases in the ground reaction forces are largely a result of the increased load
374 applied to the system, with increases in both magnitude and rate of force application influenced.

375 However, the localised changes due to carrier type in both kinetics and temporal measures indicate

376 that carrier structure has an influence beyond the magnitude of the load. Resultantly, caregivers

377 should be cautious when selecting and using such devices, as these results are based on acute

378 application only, without consideration of prolonged use. Further investigation would be merited in

379 exploring the postural changes associated with the observed alterations in ground reaction force and

380 the impact of prolonged use on the wearer.

381

382 **Key Points**

383 ▪ Carrying infants in ergonomic carriers has been said to improve the bond between caregiver
384 and child but the implications of wearer health has received limited attention.

385 ▪ Previous work has addressed maternal mothers only with no consideration for other
386 caregivers in an infant's life.

387 ▪ Results indicate that the load increases all aspects of ground reaction force, however the
388 magnitude and temporal alterations are dependent on carrier structure.

389 ▪ The structure of the carrier should be carefully considered when selecting ergonomic infant
390 carriers, as the acute alterations indicate that the structure impacts the magnitude of the

391 forces experienced during over ground walking.

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