

Gluteal Soft-Tissue Dynamics

A Damped-Oscillator and Information-Theoretic Analysis of the “Jiggle” Effect

Effects of body size, composition, tissue condition and shape

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Abstract

When a person walks or runs, the gluteal soft tissue (skin, adipose tissue and muscle) oscillates relative to the pelvis after each heel strike. We model this motion as a base-excited, viscoelastic **mass–spring–damper system** and study how its natural frequency, damping and displacement depend on body mass, body-fat fraction, muscle tone, garment support and tissue shape. We then apply **Shannon information theory** to the simulated signals: the entropy of the quantised displacement and the normalised spectral entropy describe how rich the motion is, and the Shannon–Nyquist and Shannon–Hartley results guide how it should be measured. Across the four body types the natural frequency ranged from 3.8 to 5.8 Hz and the peak running displacement from 24 to 35 mm. This extended version adds a frequency-domain analysis, a gait-resonance (cadence) study, a one-at-a-time sensitivity analysis, a 300-body Monte-Carlo population, a quantisation law for the entropy, and a simulated test of how well f_n and ζ can be recovered from sensor data. All parameter values are literature-style *assumptions* and must be validated with accelerometer or motion-capture data.

Keywords: soft-tissue vibration, damped oscillator, resonance, Shannon entropy, spectral entropy, Nyquist sampling, sports biomechanics.

Symbol	Meaning	Unit	Symbol	Meaning	Unit
M	body mass	kg	f_n, ω_n	undamped natural frequency	Hz, rad/s
BF	body-fat fraction	–	ζ	damping ratio	–
m_{eff}	effective tissue mass	kg	ω_d	damped frequency	rad/s
E_m, E_f	muscle, fat moduli (relative)	–	δ	logarithmic decrement	–
σ	shape factor	–	Q	resonance magnification	–
$z(t)$	tissue displacement	mm	H, H_s	Shannon, spectral entropy	bit
$a_b(t)$	pelvis acceleration	m/s ²	Δ	quantisation step	mm

1. Introduction

Soft-tissue motion during locomotion matters for sports-garment design, comfort, energy dissipation and impact attenuation. Soft tissue behaves like a lightly damped vibrating system excited by the vertical acceleration of the skeleton; skin-mounted accelerometer studies of leg muscle have reported damped oscillations in the same general frequency range [6,7]. Our goal is a transparent, reproducible mathematical model that explains *why* different bodies move differently, and the use of Shannon’s tools to describe the resulting signals.

Research questions. (i) How do body mass and body-fat fraction change the natural frequency and amplitude? (ii) How do muscle contraction and compression garments change them? (iii) How does tissue shape matter? (iv) How much information, in bits, is carried by the motion signal? (v) *New in this version:* how does the answer depend on running cadence, how sensitive is it to each assumed parameter, and could f_n and ζ really be recovered from realistic sensor data?

What is new. Sections 2.4–2.6 (frequency response, energy loss, assumptions), 3.2 and 3.4 (quantisation law, worked sensor budget), and 4.4–4.8 (harmonics, cadence sweep, sensitivity, population simulation, design chart), plus Section 5 (parameter recovery) and Appendix B (reproducibility). The original results are regenerated with the same model and agree with Version 1.

2. Mathematical model

2.1 Equation of motion

Let $z(t)$ be the displacement of the tissue mass relative to the pelvis and $a_b(t)$ the vertical acceleration of the pelvis. Newton's second law for the tissue, with a Kelvin–Voigt (spring + dashpot) element, gives

$$m\ddot{z} + c\dot{z} + kz = -ma_b(t) \quad (1)$$

$$\ddot{z} + 2\zeta\omega_n\dot{z} + \omega_n^2 z = -a_b(t), \quad \omega_n = \sqrt{k/m}, \quad \zeta = \frac{c}{2\sqrt{km}} \quad (2)$$

For $\zeta < 1$ the free response is a decaying oscillation at $\omega_d = \omega_n \sqrt{1-\zeta^2}$, the visible “jiggle”. With initial displacement z_0 and zero velocity,

$$z(t) = z_0 e^{-\zeta\omega_n t} \left[\cos \omega_d t + \frac{\zeta}{\sqrt{1-\zeta^2}} \sin \omega_d t \right] \quad (3)$$

Successive peaks shrink by a constant ratio, which defines the logarithmic decrement δ ; the time to fall inside a $\pm 5\%$ band is found from the envelope $e^{-\zeta\omega_n t} = 0.05$, i.e. $e^3 \approx 20$:

$$\delta = \ln \frac{z_n}{z_{n+1}} = \frac{2\pi\zeta}{\sqrt{1-\zeta^2}}, \quad t_{5\%} \approx \frac{3}{\zeta\omega_n} \quad (4)$$

2.2 Dependence on body size, composition and shape

We make the following modelling assumptions. The effective vibrating mass is a small fraction of body mass that grows with body fat, the effective modulus is a fat/muscle mixture, and stiffness follows geometric similarity:

$$m_{eff} = M(0.02 + 0.06 BF), \quad E_{eff} = (1 - BF)E_m + BF E_f, \quad k = \sigma C_0 E_{eff} m_{eff}^{1/3} \quad (5)$$

Here σ is a **shape factor** (Appendix A). The $k \propto m^{1/3}$ scaling follows from $k = EA/L$ with $A \propto m^{2/3}$ and $L \propto m^{1/3}$. The damping ratio is the mixture $\zeta = (1-BF) \cdot 0.12 + BF \cdot 0.30$, because adipose tissue dissipates more energy than muscle. The constant C_0 is calibrated so that the “Average, relaxed” case has $f_n = 5$ Hz; the fat-to-muscle modulus ratio is $E_f/E_m \approx 0.16$, muscle contraction multiplies E_m by 3.0, and a compression garment multiplies k by 1.4 and sets $\zeta = 0.32$. Substituting into ω_n gives

$$f_n = \frac{1}{2\pi} \sqrt{\frac{\sigma C_0 E_{eff}}{m_{eff}^{2/3}}} \propto m_{eff}^{-1/3} \quad (6)$$

Heavier soft-tissue mass therefore lowers the frequency (slower, larger jiggle), stiffer tissue raises it, and a larger σ (wide, flat tissue: large cross-section, short depth) raises it too.

2.3 Excitation

Each heel strike is a half-sine pulse of width 0.10 s (walking, 1.8 steps/s, peak 0.5 g) or 0.06 s (running, 2.8 steps/s, peak 2.5 g), with 8 % amplitude variability and 3 % timing jitter (standard deviations) to mimic natural stride variation. The ODE was solved numerically as a linear time-invariant system ($f_s = 1$ kHz, 20 s, first 2 s discarded). A single pulse of width w and peak A delivers an impulse of $2Aw/\pi$, so for running the impulse is ≈ 0.94 m/s and the first-swing displacement is of order impulse/ $\omega_n \approx 30$ mm before damping; the simulated 24 mm peaks are consistent with this hand estimate.

2.4 Frequency-domain description

Because the model is linear, everything about the response to *any* pelvis acceleration is contained in one transfer function. With $s = j\omega$,

$$G(j\omega) = \frac{Z(j\omega)}{A_b(j\omega)} = \frac{-1}{\omega_n^2 - \omega^2 + 2j\zeta\omega_n\omega}, \quad |G| = \frac{1}{\omega_n^2 \sqrt{(1-r^2)^2 + (2\zeta r)^2}}, \quad r = \frac{\omega}{\omega_n} \quad (7)$$

The magnification factor peaks slightly below f_n . Setting the derivative to zero gives the resonance frequency and quality factor:

$$f_r = f_n \sqrt{1 - 2\zeta^2}, \quad Q = \frac{1}{2\zeta\sqrt{1 - \zeta^2}} \approx \frac{1}{2\zeta}, \quad \Delta f_{-3dB} \approx 2\zeta f_n \quad (8)$$

The last relation is the half-power bandwidth rule used in Section 5 to estimate ζ from a measured spectrum. The force passed back to the pelvis is described by the transmissibility

$$T(r) = \sqrt{\frac{1 + (2\zeta r)^2}{(1 - r^2)^2 + (2\zeta r)^2}} \quad (9)$$

which exceeds 1 for $r < \sqrt{2}$ (the tissue amplifies the load) and falls below 1 beyond it (the tissue isolates the skeleton). A running heel-strike train has energy at the step frequency and its harmonics, $f_k = k f_{\text{step}}$; the jiggle is strongest when one of them falls close to f_r .

Table 1. Frequency-domain quantities derived from Eqs. (4), (8). “Swings to 5 %” = $t_{5\%} f_d$

Body	f_n [Hz]	ζ	f_r [Hz]	Q	Δf [Hz]	δ	swings to 5 %	energy lost per cycle
Slim	5.78	0.16	5.63	3.2	1.80	0.99	3.0	86 %
Average	5.00	0.17	4.85	3.0	1.70	1.09	2.8	89 %
Curvy	4.42	0.18	4.27	2.8	1.62	1.17	2.6	90 %
Heavier	3.78	0.20	3.64	2.6	1.48	1.25	2.4	92 %

2.5 Energy dissipation

Over one cycle the vibration energy falls by the factor $e^{-2\delta}$, so the fraction dissipated is $1 - e^{-2\delta}$. For the average body ($\zeta = 0.17$, $\delta = 1.09$) this is about 89 % per cycle: the tissue is a strong energy absorber, which is exactly why it is of interest for impact attenuation and garment design. Equivalent damping descriptors are the loss factor $\eta = 2\zeta$ (0.34) and the specific damping capacity $\psi = 4\pi\zeta$ (2.1).

2.6 Summary of assumptions

Table 2. Modelling assumptions and their weak points.

Assumption	Value used	Why / how it could be wrong
One degree of freedom, vertical	single mass on Kelvin–Voigt element	real tissue is 3-D and anisotropic
Linear stiffness	k independent of strain	tissue stiffens with strain; amplitudes of 25–35 mm may leave the linear range
Effective mass	2 % + 6 %·BF of body mass	not measured; depends on definition of the “gluteal” region
Damping ζ	0.12 (muscle) ... 0.30 (fat)	literature-style range, not fitted
Heel-strike pulse	half-sine, 0.5 g / 2.5 g peak	real pulses depend on shoe, surface and speed
Rigid pelvis input	pelvis acceleration = skeleton	ignores pelvic rotation and left–right asymmetry

3. Shannon information-theoretic analysis

3.1 Shannon entropy of the displacement

The signal is quantised in steps of $\Delta = 0.25$ mm (as a sensor would) and the probability p_i of each level is estimated from its relative frequency:

$$H(Z) = -\sum_i p_i \log_2 p_i \quad [\text{bit/sample}] \quad (10)$$

A larger H means more distinct displacement levels are used with comparable probability: the motion is richer and needs more bits per sample to encode. A rigid, almost non-moving tissue gives $H \rightarrow 0$.

3.2 Entropy depends on the quantiser: a scaling law

H is not a property of the motion alone; it also depends on Δ . When Δ is small compared with the signal spread, the quantised entropy approaches the differential entropy $h(Z)$ of the continuous signal minus $\log_2 \Delta$:

$$H(Z_\Delta) \approx h(Z) - \log_2 \Delta \quad (11)$$

Each halving of Δ therefore adds exactly one bit. Section 4.9 verifies this on the simulated signal, where $H + \log_2 \Delta$ is constant at ≈ 4.9 bit (Δ in mm). The practical consequence is that entropy values from different sensors are comparable only if Δ is quoted; the 0.25 mm used throughout this paper is a stated convention, not a physical constant.

3.3 Spectral entropy

From the power spectral density $S(f)$ (Welch method, 2048-sample segments, 0.5–50 Hz) we form a probability distribution over the N frequency bins and normalise by $\log_2 N$ so that $H_s \in [0, 1]$:

$$H_s = -\frac{1}{\log_2 N} \sum_{k=1}^N \hat{P}_k \log_2 \hat{P}_k, \quad \hat{P}_k = \frac{S(f_k)}{\sum_j S(f_j)} \quad (12)$$

$H_s \approx 0$ means a single dominant resonance; $H_s \approx 1$ means a flat, noise-like spectrum. Its value depends on where the natural frequency sits relative to the broad spectrum of the heel-strike pulses, not on damping alone.

3.4 Sampling, capacity and a worked sensor budget

To record the effect without aliasing, the Shannon–Nyquist theorem requires $f_s \geq 2f_{\max}$. Because the jiggle has content up to about 20–30 Hz, a sensor at ≥ 100 Hz is safe and we recommend 200 Hz. The Shannon–Hartley theorem gives the maximum error-free information rate of a channel with bandwidth B and signal-to-noise ratio S/N :

$$f_s \geq 2f_{\max}, \quad C = B \log_2 \left(1 + \frac{S}{N} \right) \quad (13)$$

Worked example. Take $f_s = 200$ Hz, a displacement range of ± 40 mm and $\Delta = 0.25$ mm. The number of levels is $80/0.25 = 320$, so an ADC needs $\lceil \log_2 320 \rceil = 9$ bit. The sample entropy of running is $H = 6.96$ bit, so the naive data rate is $6.96 \times 200 \approx 1391$ bit/s. A channel of bandwidth $B = 50$ Hz and SNR = 30 dB ($S/N = 1000$) has capacity $50 \cdot \log_2(1001) \approx 498$ bit/s. The naive figure is nearly three times larger than the capacity because consecutive samples of a narrow-band oscillation are strongly correlated: the *entropy rate* of the process is much smaller than Hf_s . Sample entropy is therefore a useful *descriptor*, but it overstates the information actually conveyed, and it should not be read as a bit-rate.

4. Results

4.1 Body size and composition

Running, relaxed, average shape ($\sigma = 1$).

Table 3. Simulated results for four body types.

Type	M [kg]	BF [%]	m_{eff} [kg]	f_n [Hz]	ζ	Peak [mm]	RMS [mm]	$t_{5\%}$ [ms]	H [bit]	H_s
Slim	55	20	1.76	5.78	0.16	23.6	8.7	530	6.87	0.37
Average	65	28	2.39	5.00	0.17	24.0	8.9	561	6.96	0.41
Curvy	75	35	3.07	4.42	0.18	25.5	9.3	591	6.95	0.41
Heavier	95	42	4.29	3.78	0.20	34.7	12.6	645	7.37	0.32

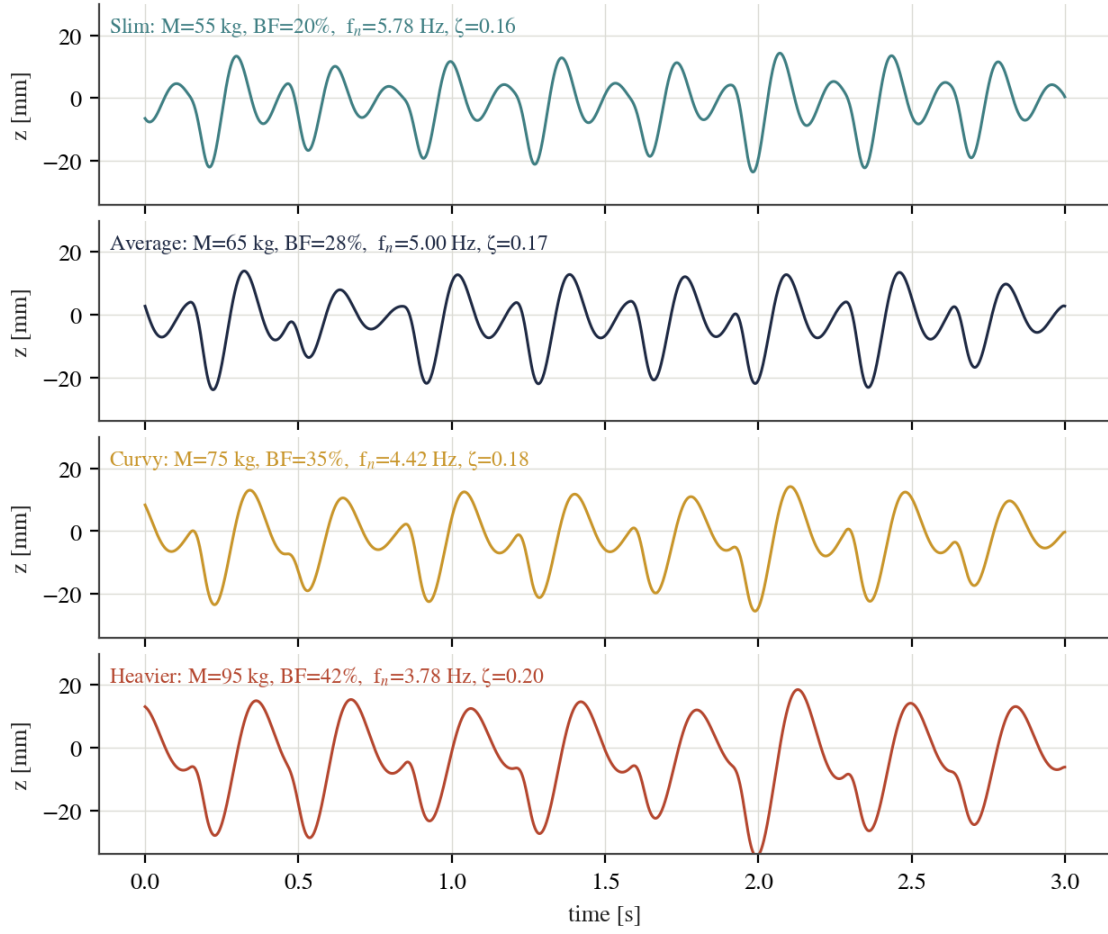


Figure 1. Simulated tissue displacement $z(t)$ for the four body types (3 s excerpt, steady state).

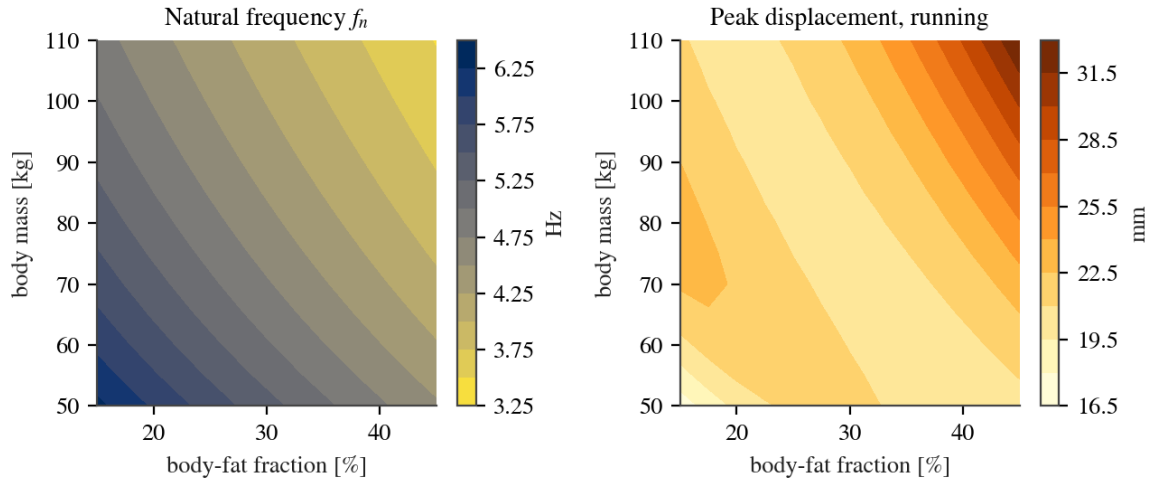


Figure 2. Parameter sweep over body mass and body-fat fraction: natural frequency (left) and peak running displacement (right; periodic-train simulation without stride variability, hence lower than Table 3).

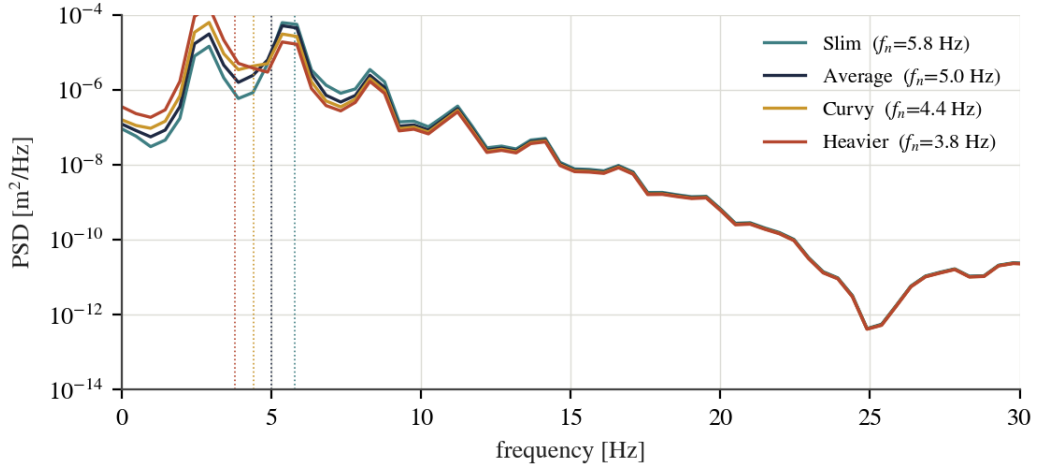


Figure 3. Power spectral density of the displacement. Dotted lines mark each case's natural frequency; the dominant peaks sit near them.

4.2 Condition

Average body: 65 kg, 28 % body fat.

Table 4. Effect of activity, muscle contraction and compression garment.

Condition	f_n [Hz]	ζ	Peak [mm]	RMS [mm]	$t_{5\%}$ [ms]	H [bit]	H_s
Walking, relaxed	5.00	0.17	6.8	2.2	561	4.83	0.47
Running, relaxed	5.00	0.17	24.0	8.9	561	6.96	0.41
Running, muscle contracted	8.49	0.17	12.5	3.8	330	5.57	0.51
Running, compression garment	5.91	0.32	17.2	5.1	252	5.56	0.45

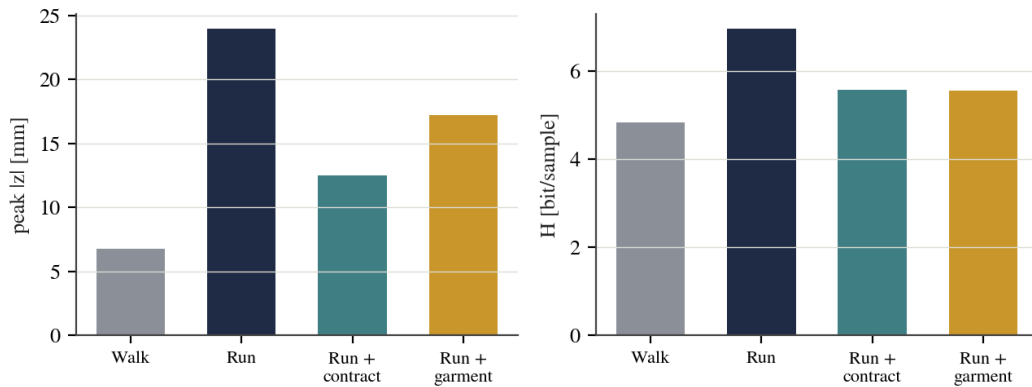


Figure 4. Peak displacement (left) and Shannon entropy (right) by condition.

4.3 Shape

Table 5. Effect of tissue shape factor σ (average body, running).

Shape	σ	f_n [Hz]	Peak [mm]	RMS [mm]	H [bit]	H_s
Flat / wide	1.15	5.36	23.7	8.9	6.90	0.38
Round	1.00	5.00	24.0	8.9	6.96	0.41
Square	1.10	5.24	24.0	8.9	6.92	0.39
High / full	0.88	4.69	24.5	9.0	6.97	0.42

4.4 Frequency response and gait harmonics

Figure 5 plots Eqs. (7) and (9) for the four bodies. All four curves peak with a magnification of about 2.7–3.4 and are separated mainly in frequency; the heavier the tissue, the further left the resonance sits. The dashed lines mark the harmonics of a 2.8 steps/s run. The slim body’s resonance (5.7 Hz) lies almost exactly on the second harmonic (5.6 Hz), whereas the heavier body (3.7 Hz) sits between the first and second harmonics. Phase portraits of the same four simulations (Figure 6) show the corresponding trajectories: tighter, faster loops for the stiff tissue and larger, slower loops for the heavy one.

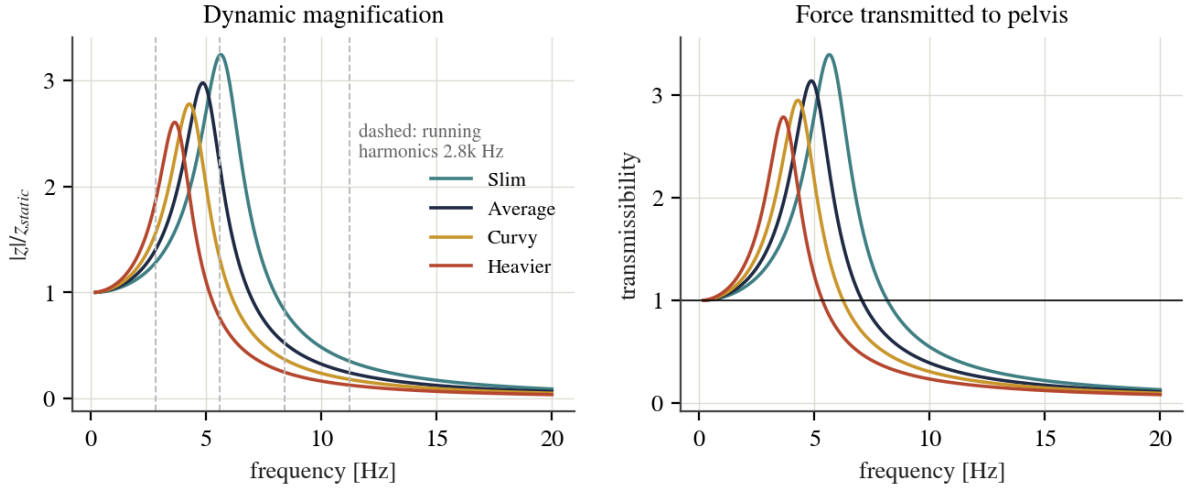


Figure 5. Dynamic magnification (left) and transmissibility (right) for the four body types.

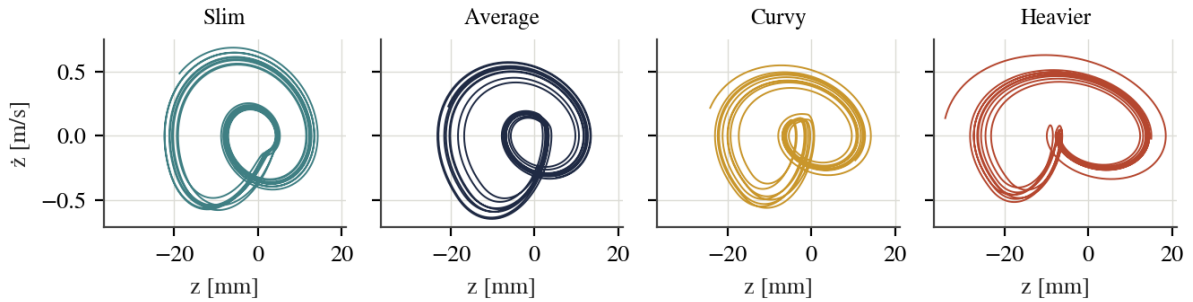


Figure 6. Phase portraits (displacement against velocity) from the running simulations, 3 s each.

4.5 Cadence sweep: the gait-resonance effect

Table 3 suggests “heavier means bigger jiggle”, but amplitude also depends on cadence. Figure 7 sweeps the step frequency of a periodic running train from 1.6 to 3.6 steps/s. Peaks occur where a harmonic of the step frequency meets the resonance: at 2.87 steps/s for the slim body, 2.46 for the average, and 2.17 for the curvy body, in each case close to $f_n/2$ (2.89, 2.50 and 2.21 Hz), i.e. the second harmonic. The heavier body’s first-harmonic resonance (3.78 steps/s) lies at the edge of the plotted range, which is why its curve is still rising. At the nominal 2.8 steps/s the four periodic-train peaks are close to each other for the first three bodies, so the larger amplitude of the “Curvy” case in Table 3 is partly a matter of where the nominal cadence happens to fall. **The mass trend is therefore a tendency, not a law:** a body can reduce its jiggle by shifting cadence away from a resonance, just as a runner’s own gait changes in practice.

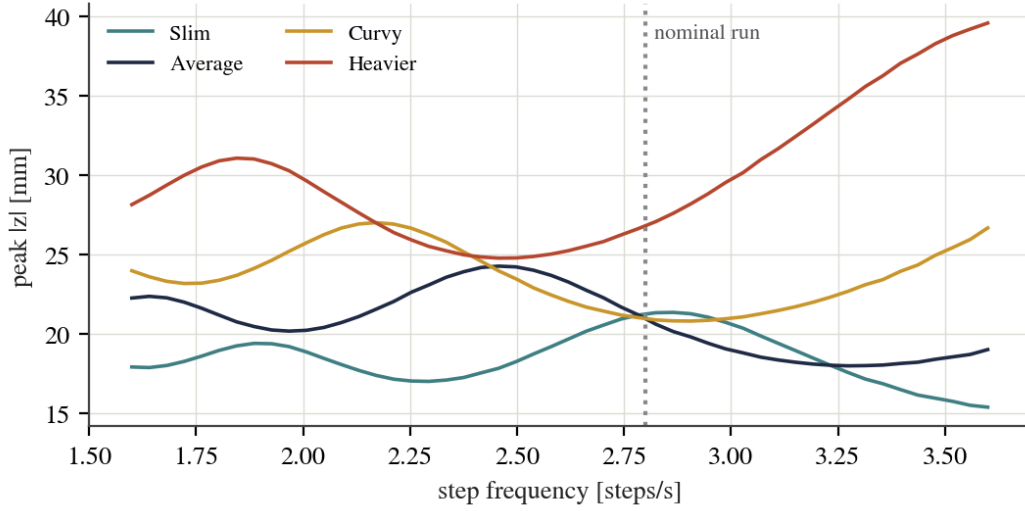


Figure 7. Peak displacement versus step frequency (deterministic periodic pulses of 2.5 g, 0.06 s). Local maxima occur where a gait harmonic coincides with the natural frequency.

4.6 Sensitivity to the assumed parameters

Because most inputs are assumptions, we vary each by $\pm 20\%$ around the average running case (baseline periodic-train peak 21.0 mm) and record the change in peak displacement (Figure 8). The model is linear in the pulse size, so the impact peak dominates trivially (± 4.2 mm). Among tissue properties, damping ratio and body-fat fraction matter most (about ± 1 mm), then body mass, muscle modulus and shape factor ($\approx \pm 0.2$ – 0.5 mm), and the fat modulus almost not at all ($\approx \pm 0.06$ mm). Two points follow. First, the ranking of ζ among the tissue parameters says that measuring damping is a priority for validation. Second, the asymmetries (e.g. body mass $+20\%$ lowers the periodic-train peak by 0.4 mm but -20% raises it by 0.5 mm) are the resonance effect of Section 4.5 showing up again: a parameter change can move the system toward or away from a harmonic.

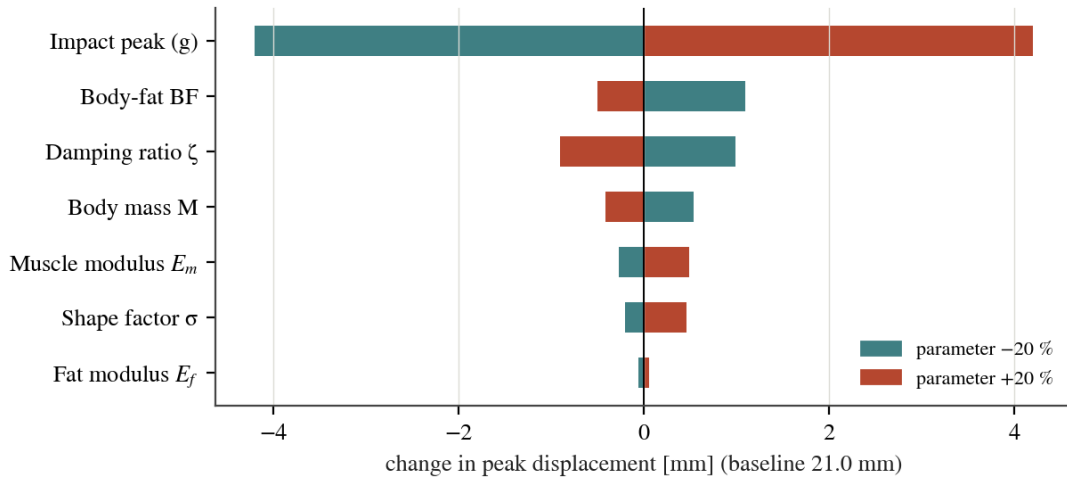


Figure 8. One-at-a-time sensitivity of the peak running displacement to $\pm 20\%$ changes in each parameter.

4.7 A simulated population

To go beyond four hand-picked bodies we drew 300 virtual participants: M uniform 50–110 kg, BF uniform 15–45 %, $\sigma \sim N(1, 0.08^2)$, with 50 % relaxed, 25 % contracted and 25 % garment-supported tissue. The natural frequency has mean 5.6 Hz and 5–95 % range 3.9–8.7 Hz; the periodic-train peak displacement has mean 18 mm and 5–95 % range 11–25 mm. Among relaxed bodies, log peak against log f_n has correlation -0.68 and slope -0.49 , i.e. peak $\propto f_n^{-0.5}$ as a rule of thumb, and the correlation between peak and body mass is only 0.46: **frequency is a better predictor of amplitude than mass is**. Tone and garment move bodies to the right of the cloud (stiffer, smaller motion).

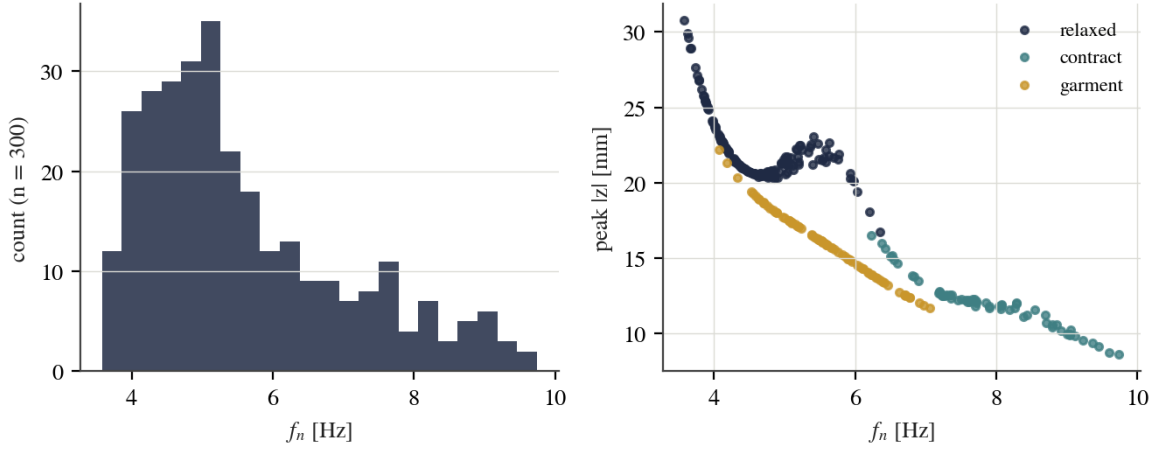


Figure 9. Monte-Carlo population ($n = 300$). Left: distribution of natural frequency. Right: peak displacement against natural frequency, coloured by tissue condition.

4.8 Design chart

Figure 10 places every case on the (f_n, ζ) plane with contours of the 5 % settling time $t_{5\%} = 3/(\zeta\omega_n)$. It makes the garment result visible at a glance: contracted muscle moves a body right (faster), a garment moves it right *and* up (faster and better damped), and only the latter brings settling below 300 ms. A garment designer can read off the damping needed to reach a target settling time at a given stiffness.

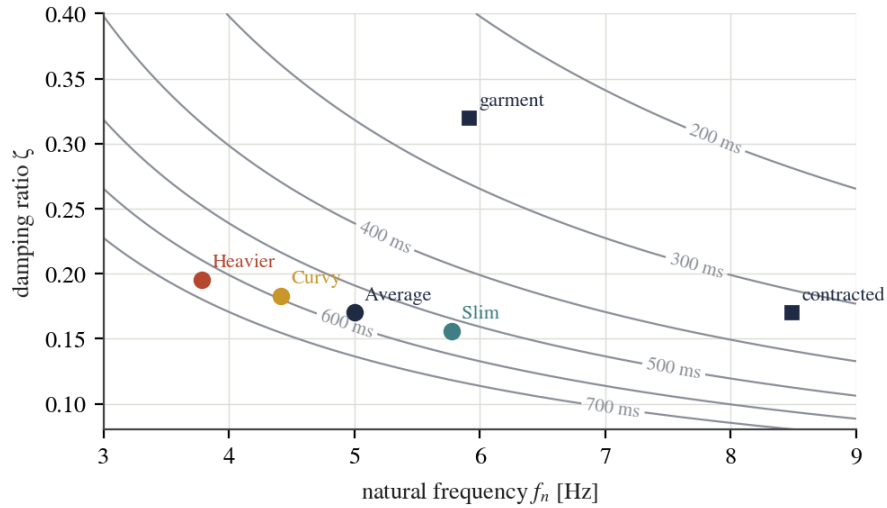


Figure 10. Settling-time contours on the natural frequency–damping plane. Circles: relaxed running; squares: contracted muscle and garment.

4.9 Entropy against quantisation step

Figure 11 confirms Eq. (11). Left: H falls steadily as Δ grows, and the dotted line marks the 0.25 mm used here. Right: after adding $\log_2 \Delta$ the curve is flat for small Δ , so the quantity $h(Z)$ is a property of the motion itself, while the departure at large Δ (a few mm) shows when the quantiser becomes too coarse to resolve the signal.

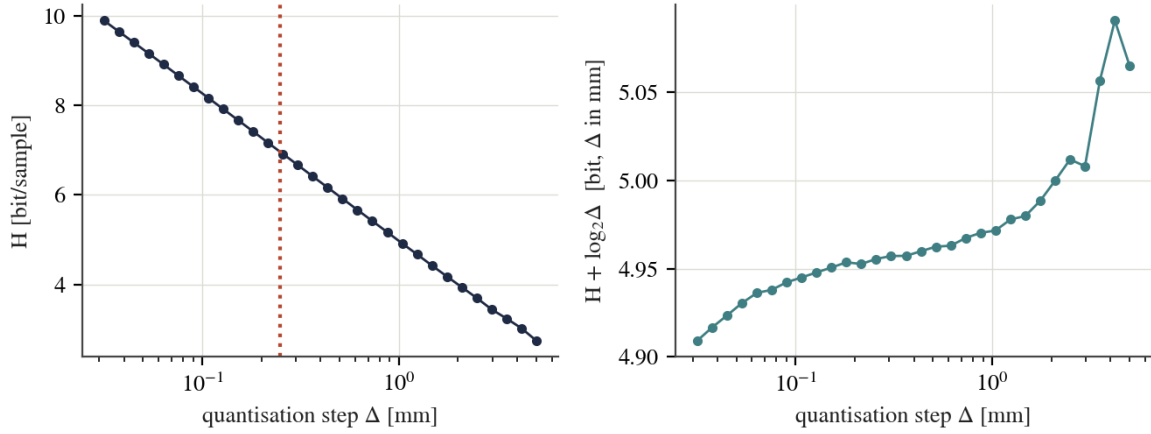


Figure 11. Sample entropy of the average running signal against quantisation step (left) and the invariant $H + \log_2 \Delta$ (right).

5. Can f_n and ζ be recovered from sensor data?

The validation plan of Section 7 rests on estimating f_n and ζ from two accelerometers. We tested this on simulation. For the average body, 60 s of running was generated with stride variability; white sensor noise was added to both the pelvis acceleration a and the tissue displacement z at SNR = 40 ... 5 dB; the transfer function was estimated as $\hat{G} = P_{az}/P_{aa}$ (Welch, 2048 samples) and Eq. (7) was fitted over 1.5–14 Hz by coherence-weighted least squares. 25 trials were run at each SNR.

The estimator is **precise but slightly biased**. The median f_n is 4.92 Hz at 40 dB and 4.92 Hz at 5 dB (true 5.00 Hz; inter-quartile range ≤ 0.02 Hz), so noise barely matters down to 5 dB. The damping is overestimated, $\zeta \approx 0.19$ –0.20 against a true 0.17, a bias of about 14–17 %. The likely cause is the 1 Hz spectral resolution of the Welch segments and leakage across the harmonic structure of the periodic input, not sensor noise. Two practical conclusions: frequency can be trusted to within about 2 %, but damping needs longer records, finer resolution (segments ≥ 4 s) or a time-domain fit to isolated decays before the number is quoted. Real sensors add errors that are not modelled here (skin-sliding, sensor mass loading, soft-tissue artefact in the pelvis marker).

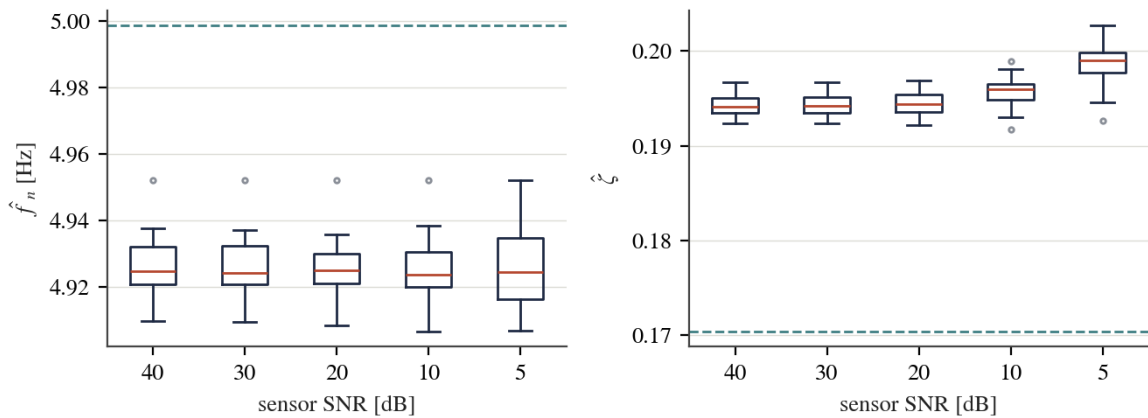


Figure 12. Recovered natural frequency (left) and damping ratio (right) against sensor SNR; boxes are 25 trials, dashed line is the true value.

6. Discussion

Mass and fat. Increasing soft-tissue mass lowers f_n ($\propto m^{-1/3}$; 5.8 $\rightarrow$ 3.8 Hz from Slim to Heavier) and increases displacement (24 $\rightarrow$ 35 mm), because the same pelvis acceleration drives a larger, softer mass. Body fat raises the damping ratio, but the lower f_n more than cancels this in absolute time, so the settling time still grows (529 $\rightarrow$ 645 ms). Sections 4.5–4.7 add an important caveat: the amplitude trend is only statistical, and cadence relative to resonance can reverse it for a given body.

Muscle tone and garments. Muscle tone raises stiffness, so f_n rises to 8.5 Hz and amplitude roughly halves (12.5 mm), the largest reduction of the tested conditions. Compression garments increase both k and ζ , giving the fastest settling (252 ms) and a 28 % amplitude reduction. Shape acts through σ and mainly shifts f_n (4.7–5.4 Hz), with a small effect on amplitude.

Information view. Shannon entropy tracks amplitude: larger motion occupies more quantisation levels ($H = 6.96$ bit/sample running vs 4.83 walking, and 5.56–5.57 with contraction or a garment). Spectral entropy separates the character of the motion from its size: H_s is lowest for the heaviest body (0.32), whose low f_n concentrates energy near one resonance, and highest for the stiffened tissue (0.51), whose higher f_n overlaps more of the broad pulse spectrum. Together H and H_s give a compact two-number description of “how much” and “how complex” the jiggle is, with the caveat of Section 3.2 that H is only meaningful at a stated Δ .

7. Limitations and validation plan

- The model is 1-D and linear; real tissue is anisotropic and non-linear (stiffness depends on strain).
- Moduli, damping ratios, C_0 and σ values are assumed, not measured. The tables show trends, not clinical facts.
- Left–right asymmetry, skin tension and pelvic motion in other axes are ignored.
- The recovery test of Section 5 uses white noise and a perfect pelvis reference; real data will be worse.

Table 6. Proposed experimental protocol.

Item	Proposal
Sensors	two IMUs (± 16 g, ≥ 200 Hz): one on the sacrum, one on the gluteal belly, mass < 5 g, double-sided tape plus elastic strap; time-synchronised
Alternative	marker-based motion capture, ≥ 200 Hz, pelvis cluster plus skin markers
Tasks	treadmill walking (1.4 m/s) and running (2.5, 3.0, 3.5 m/s), 60 s each; relaxed, then voluntary gluteal contraction, then garment
Outputs	f_n and ζ by transfer-function fit and by half-power bandwidth ($\zeta \approx \Delta f/2f_n$); peak, RMS, H at $\Delta = 0.25$ mm, H_s
Analysis	regress f_n and ζ on mass, body-fat % and σ (superellipse fit, Appendix A); report cadence for every record
Sample	≥ 10 participants per predictor (about 30–40 for three predictors); informed consent, university ethics approval, data anonymised

8. Conclusion

Short answer: yes. In this model, the gluteal soft tissue is capable of jiggling, and it can be treated as a lightly damped vibrating system, a mass on a spring with a shock absorber. It is not a rigid block, and it is not a heavily damped one that would simply sag and stop.

1. Underdamped. Every simulated case has ζ between 0.16 and 0.32, well below 1; after a push it overshoots and swings back and forth before it settles. That swinging is the “jiggle”.

2. Visible. The natural frequency is about 3.8–5.8 Hz (8.5 Hz when the muscle is contracted), so it oscillates roughly 4–6 times per second, and it takes about 0.25–0.65 s to calm down, about 1.5 to 3 full swings after each heel strike (Figure 13).

3. Measurable. Peak displacement is about 7 mm when walking and 24–35 mm when running (average body: 24 mm).

4. Information measures agree. The displacement signal carries 4.8–7.4 bit per sample at $\Delta = 0.25$ mm, and its spectrum has a clear peak near the natural frequency, as expected for a resonating system.

5. New in this version. The amplitude depends on where the gait harmonics fall relative to resonance, so mass alone is a weak predictor ($r \approx 0.5$) while natural frequency is a better one ($r \approx -0.7$); damping is the tissue parameter whose uncertainty matters most; and frequency can be recovered from sensors to within a couple of percent, while damping needs more careful estimation.

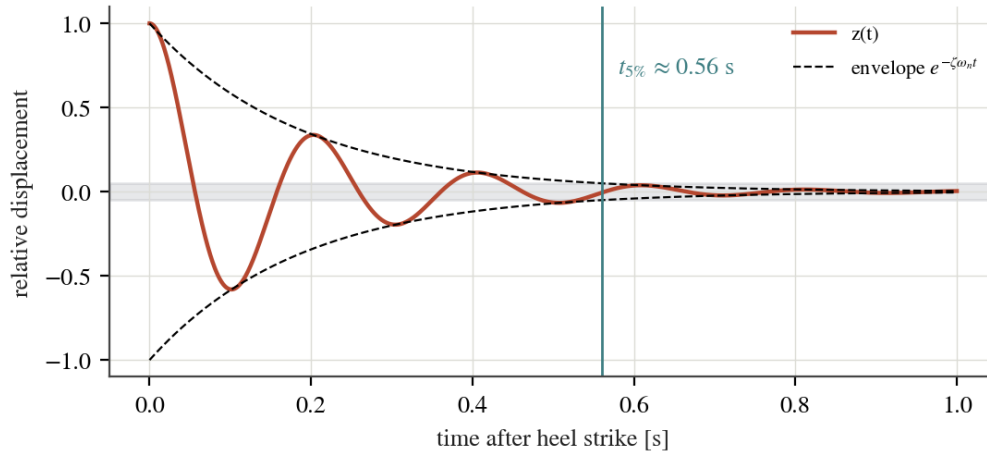


Figure 13. The jiggle as a damped oscillation: the free response of the average case settles within about half a second.

Limits. The model is simple (one direction, linear) and the stiffness and damping values are assumptions. The results show *trends* and a physically sensible explanation, not clinical facts. The next step is to measure real people with accelerometers or motion capture and check whether the real f_n and ζ match the predictions.

Appendix A. Geometry of gluteal tissue shape

Section 2.2 introduced the shape factor σ as a single number. This appendix gives it a geometric meaning, so that each tissue shape in Table 5 corresponds to a concrete, drawable cross-section. We represent the tissue in a mid-sagittal (side-view) cross-section as a **superellipse** (Lamé curve) with half-width a , half-depth b and shape exponent n :

$$\left|\frac{x}{a}\right|^n + \left|\frac{y}{b}\right|^n = 1 \quad (14)$$

For $n = 2$ this is an ellipse (round), for $n > 2$ the outline becomes boxier (square), and for $a/b > 1$ or < 1 it is wide and flat or tall and full. The enclosed area has a closed form:

$$A = 4ab \frac{\Gamma(1+1/n)^2}{\Gamma(1+2/n)} = 4ab g(n) \quad (15)$$

Because $k = EA/L$, stiffness depends on the cross-section area A and the tissue depth $L = 2b$. For a fixed tissue mass the dimensionless ratio A/L^2 controls the geometry, so we define the shape factor relative to the round reference ($a = b = 1$, $n = 2$, $g(2) = \pi/4$):

$$\sigma = \frac{A/L^2}{(A/L^2)_{\text{round}}} = \frac{a}{b} \cdot \frac{g(n)}{g(2)} \quad (16)$$

With the parameters in Figure 14 this reproduces the σ values of Table 5 (1.15, 1.00, 1.10, 0.88) and therefore $f_n = 5.0 \cdot \sqrt{\sigma}$ Hz = 5.36, 5.00, 5.24 and 4.69 Hz, in agreement with the simulation. Wide, shallow tissue has a large σ (stiffer, faster); tall, deep tissue has a small σ (softer, slower).

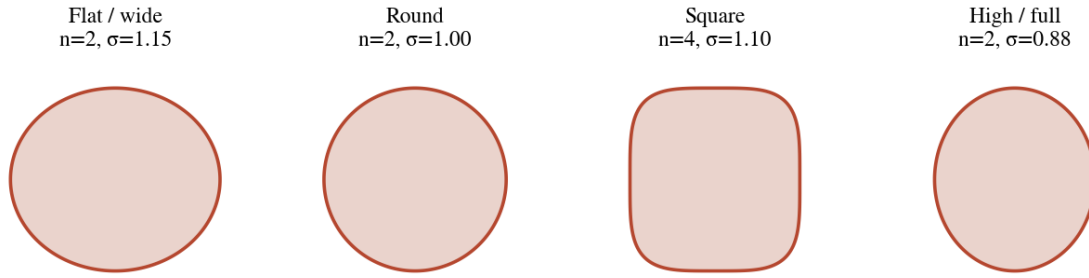


Figure 14. Superellipse cross-sections for the four shape classes of Table 5, with exponent n and resulting shape factor σ .

A.1 Geometric similarity and body size

The scaling $k \propto m^{1/3}$ assumes geometric similarity: all lengths grow with the cube root of tissue mass and areas with its two-thirds power. Figure 15 shows the same round outline scaled to the effective masses of Table 3: from Slim to Heavier the linear size grows by $(4.29/1.76)^{1/3} \approx 1.35$ while the area grows by ≈ 1.82 . Stiffness therefore rises more slowly than mass, which is why f_n falls with body size.

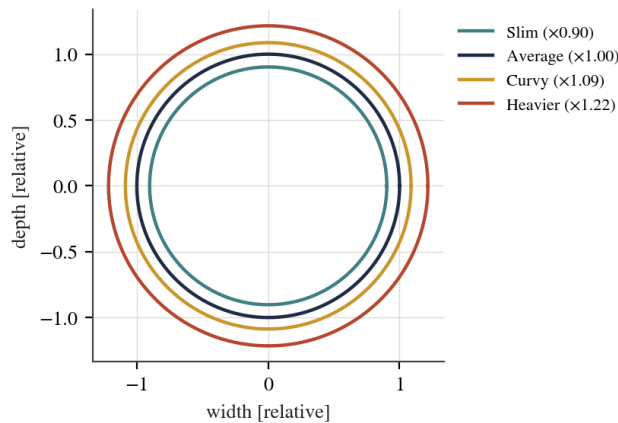


Figure 15. Geometric similarity: the round cross-section scaled by $(m_{\text{eff}}/2.39 \text{ kg})^{1/3}$.

A.2 Shape space

Figure 16 (left) maps σ over the $(n, a/b)$ plane with the four shapes marked. Any measured outline can be fitted to a superellipse and placed on this map. Since $f_n \propto \sqrt{\sigma}$, a 15 % change in σ changes the natural frequency by only about 7 %, consistent with the small amplitude differences in Table 5. In the validation study, outlines can be extracted from side-view photographs or marker sets (with informed consent), fitted by least squares for (a, b, n) , and then used to compute σ for each participant.

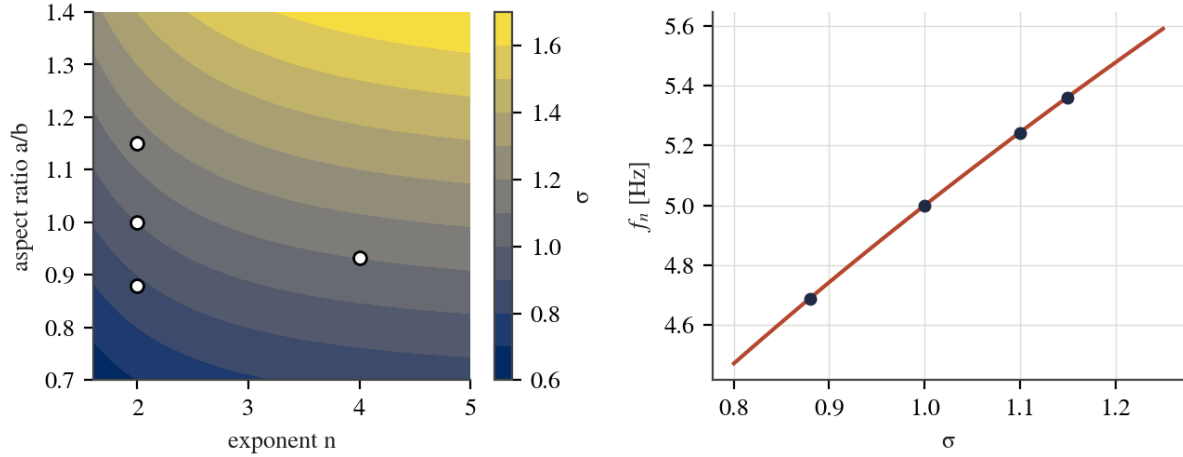


Figure 16. Left: shape factor σ over exponent n and aspect ratio a/b . Right: natural frequency against σ (dots: simulated shapes).

Appendix B. Reproducibility

Table 7. Parameters needed to reproduce the numbers.

Quantity	Value
Calibration	f_n (65 kg, 28 %, $\sigma = 1$, relaxed) = 5.00 Hz
E_f/E_m	0.159 (from the Slim/Average frequency ratio of Version 1)
Contraction, garment	$E_m \times 3.0$; $k \times 1.4$ and $\zeta = 0.32$
Solver	linear system simulation (SciPy <code>lsim</code>), 1 kHz, 20 s, first 2 s discarded, seed 1
Entropy	$\Delta = 0.25$ mm; Welch 2048 samples, 0.5–50 Hz
Sweeps, tornado, population	periodic pulse train without jitter, 6–8 s, so peaks are ≈ 12 % below the stochastic values of Table 3
Population	300 bodies, generator seed 7

References

- [1] C. E. Shannon, “A Mathematical Theory of Communication,” *Bell System Technical Journal*, 27, 1948.
- [2] C. E. Shannon, “Communication in the Presence of Noise,” *Proc. IRE*, 37(1), 1949.
- [3] S. S. Rao, *Mechanical Vibrations*, Pearson.
- [4] Y. C. Fung, *Biomechanics: Mechanical Properties of Living Tissues*, Springer.
- [5] T. M. Cover and J. A. Thomas, *Elements of Information Theory*, Wiley.
- [6] J. M. Wakeling and B. M. Nigg, “Soft-tissue vibrations in the quadriceps measured with skin mounted transducers,” *J. Biomechanics*, 34(4), 2001.
- [7] B. M. Nigg and J. M. Wakeling, “Impact forces and muscle tuning: a new paradigm,” *Exercise and Sport Sciences Reviews*, 29(1), 2001.
- [8] P. D. Welch, “The use of fast Fourier transform for the estimation of power spectra,” *IEEE Trans. Audio Electroacoust.*, 15(2), 1967.
- [9] D. J. Inman, *Engineering Vibration*, Pearson.

Citations [6]–[9] were added in this version from memory; check volume and page details before submission.