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Digital image correlation techniques for strain measurement in a variety of biomechanical test models

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Purpose: Previous biomechanical studies have estimated the strains of bone and bone substitutes using strain gages. However, applying strain gages to biological samples can be difficult, and data collection is limited to a small area under the strain gage. The purpose of this study was to compare digital image correlation (DIC) strain measurements to those obtained from strain gages in order to assess the applicability of DIC technology to common biomechanical testing scenarios. *Methods*: Compression and bending tests were conducted on aluminum alloy, polyurethane foam, and laminated polyurethane foam specimens. Simplified single-legged stance loads were applied to composite and cadaveric femurs. *Results*: Results showed no significant differences in principal strain values (or variances) between strain gage and DIC measurements on the aluminum alloy and laminated polyurethane foam specimens. There were significant differences between the principal strain measurements of the non-laminated polyurethane foam specimens, but the deviation from theoretical results was similar for both measurement techniques. DIC and strain gage data matched well in 83.3% of all measurements in composite femur models and in 58.3% of data points in cadaveric specimens. Increased variation in cadaveric data was expected, and is associated with the well-documented variability of strain gage analysis on hard tissues as a function of bone temperature, hydration, gage protection, and other factors specific to cadaveric biomechanical testing. *Conclusions*: DIC techniques provide similar results to those obtained from strain gages across standard and anatomical specimens while providing the advantages of reduced specimen preparation time and full-field data analysis.

Key words: *finite element analysis, biomechanics, digital image correlation (DIC), strain gage*

1. Introduction

Strain measurement is a common tool used to evaluate the mechanical behavior of bone *in vitro*. Historically, bone strains have been measured by strain gages directly bonded to bone and have become the "gold standard" in biomechanics laboratories since their introduction in the late 1950s [20]. Despite their widespread application, strain gages require detailed surface preparation and provide strain results only at the gage application site. This lack of full-field strain information limits the interpretation of experimental results.

More recently, the digital image correlation (DIC) technique has been introduced into the field of biomechanics as a full-field alternative approach to strain gage application [8]–[10], [12]–[15], [24], [25], [27], [29]. DIC is an optical full-field technique for non-contact, three-dimensional deformation measurements. The DIC technique uses a high contrast speckle pattern applied onto the surface of a sample which is recorded by digital, high resolution charge-coupled device (CCD) cameras during loading. The entire field of view is divided into a number of unique regions, called facets, and each region encompasses a subset of image pixels. DIC software tracks the characteristic features of the facets in the speckle pattern during loading to provide a progressive measure of surface deformation. Surface deformations under load are compared to unloaded measurements to

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calculate strain. Strain averaged over a set of facets results in a data point similar to that of strain gage reading.

When first introduced, DIC was commonly used in fields requiring measurement of relatively large strain magnitudes. Advances in digital imaging has increased digital camera resolution and subsequently greater precision in strain measurement [12]. In the past decade a number of commercially available DIC systems has increased, lending the ability for the technology's utilization across more fields and to smaller research institutions. The utilization of DIC within the field of biomechanics for strain measurement and specialized applications is relatively new. DIC is well-suited for biological applications as it can be used to accurately determine surface strain in inhomogeneous, anisotropic, non-linear materials such as bone. Among many applications in biomechanics, DIC has been used to determine bone surface strains during loading and after cortical bone adaptation in mouse tibiae [27], as a validation tool for a finite element (FE) models [9], [10], [13]–[15] to measure surface strains on a composite hemi-pelvis [8], and to describe relative micromotion between implanted prosthesis and neighboring bone [25], [26]. DIC yields full-field strain data on the surface of a test specimen, provides detailed resolution of strain magnitudes and directions, and is tolerant of complex specimen geometry.

While the DIC methodology is promising and has been used in an increasing number of biomechanical laboratory experiments, direct comparison between DIC and "gold standard" strain gage results on a broad range of common biomechanical test models is limited, with many available studies drawing comparisons with only composite specimens or using two-dimensional DIC methods [2], [4], [11], [19], [29]. The aim of the study was to compare DIC strain measurement to the use of strain gage data and numerical simulations across a range of substrate materials commonly used in biomechanical studies, including foam, composite sheet and both composite and cadaveric femur specimens implanted with total hip stems. The role of the substrate material on the accuracy of DIC strain measurements was investigated.

2. Materials and methods

Material selection and machining

Five materials were evaluated in this study in order to fully investigate the behavior of the DIC strain measurement system in common biomechanical testing scenarios: 2024-T4 aluminum, solid rigid polyurethane foam (Pacific Research Laboratories, Vashon, WA) with a density of 0.48 g/cc, a composite foam block comprised of a low-density (0.48 g/cc) solid rigid polyurethane foam laminated with short-fiber-filled epoxy sheets (1.64 g/cc) (Pacific Research Laboratories, Vashon, WA), fourth generation composite femurs (Sawbones, Vashon, WA), and fresh-frozen cadaveric femurs. The aluminum specimens were selected for their accepted and predictable behavior and similarity to other high-strength implant materials that are commonly tested, such as titanium and cobalt chrome alloys. The non-laminated polyurethane foam specimens closely simulate cancellous bone and are used frequently in biomechanical testing scenarios [1], [3], [21]. The composite femurs were used for their reduced interspecimen variability and ability to simulate the mechanical behavior of natural bone in compression and bending [16]. As a final comparison, cadaveric femurs provide the most realistic physiological scenario. For strain evaluation in the femur specimens, total hip arthroplasty uncemented femoral stems (Echo Bi-Metric, Biomet, Inc., Warsaw, IN) were implanted into five composite femurs and four fresh-frozen cadaveric femurs using standard surgical instrumentation. Specimen dimensions are provided in Table 1 and cadaveric femur demographics are provided in Table 2. Due to poor bone quality in one cadaveric specimen, one implant failed to securely interface with the surrounding cancellous bone and was not included in analysis.

Table 1. Experimental specimen dimensions (excluding composite and cadaveric femurs)

Material	Loading	n	L_1 (mm)	L_2 (mm)	L_3 (mm)
Aluminum alloy	Bending	5	6.3 ± 0.05	25.5 ± 0.13	304.8 ± 0.01
Polyurethane foam	Compression	5	50.0 ± 0.11	38.3 ± 0.05	113.8 ± 0.16
Polyurethane foam	Bending	5	19.2 ± 0.28	38.14 ± 0.03	317.5 ± 0.01
Laminated polyurethane foam	Compression	5	120.5 ± 0.08	32.0 ± 0.16	43.3 ± 0.11
Laminated polyurethane foam	Bending	5	170.5 ± 0.01	37.8 ± 0.25	22.4 ± 0.50

Table 2. Cadaveric specimen demographics

	Age	BMI	Gender
Bone 1	74	29	Male
Bone 2	74	19	Male
Bone 3	63	20	Male
Bone 4	63	23	Male

Mechanical testing – Bending

A Model 010-B cantilever calibration fixture (Vishay MicroMeasurements, Raleigh, NC) was used to precisely and repeatedly apply 12.7 mm displacements to the 2024-T4 aluminum alloy beams 25.4 mm from the free end of the cantilever beam (Fig. 1). Similarly, a custom fixture was developed to apply 9.81 N loads to the polyurethane foam beams at a location 25.4 mm from the free end of the foam beams. Small loads were necessary in bending tests of the polyurethane foam beams to avoid specimen failure and to achieve a close match between aluminum and foam bending specimens. An identical fixture was used to apply 137.9 N of load to the laminated polyurethane foam beams.

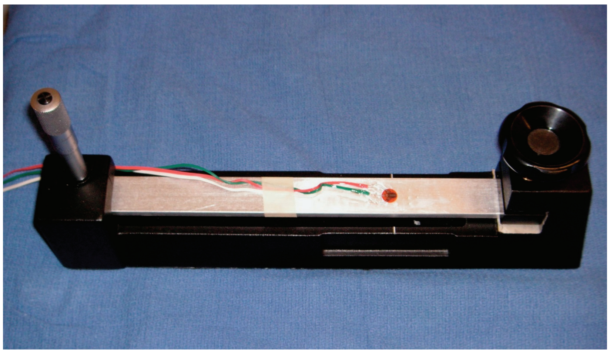

Fig. 1. Cantilever bending setup for applying a fixed deformation to an aluminum test specimen

Mechanical testing – Compression

An electrodynamic axial-torsional materials testing machine (ElectroPuls E10000 A/T, Instron Corp, Norwood, MA) was used to ramp compressive loads at a rate of 60 N/s to 2.5 kN and 5.0 kN, respectively, to the polyurethane foam and laminated polyurethane foam blocks. Loading was directed through a loading platen rigidly fixed superiorly to the testing actuator and inferiorly to the load frame base.

Mechanical testing – Single-legged Stance

Cadaveric and composite femurs were distally potted to a depth of 80 mm in polyester resin and aligned in 12 degrees of anatomic varus (Fig. 2). This setup has been described more fully in previous literature [25] and is consistent with the scenario of a simplified single-legged stance model [18], which incorporates simultaneous compression and bending forces and results in a more physiological loading scenario than isolated compression or bending loading regimens. A joint reaction force was applied to the head of the femoral component at a rate 60 N/s to a peak compressive force of 600 N at which strain data was simultaneously recorded in both DIC and strain gage systems. Trials were repeated five times with five-minute rest periods between each trial.

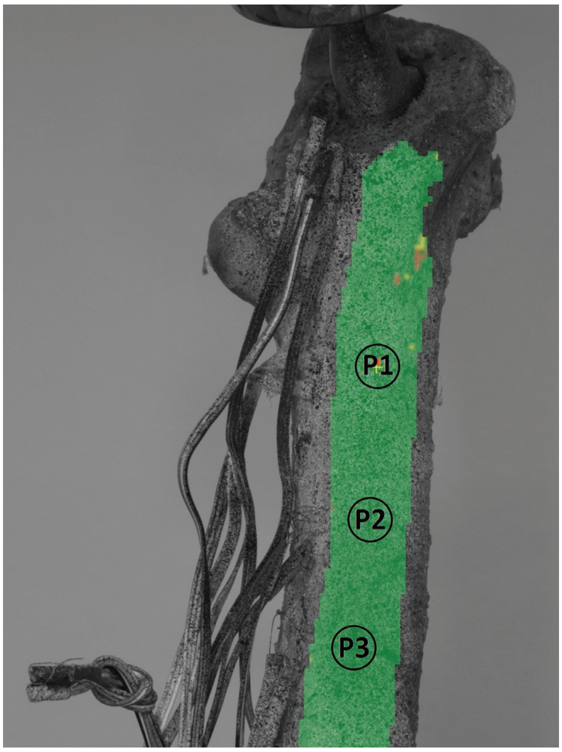

Fig. 2. Cadaveric femur specimen mounted for compression testing with speckle paint and strain gage measurement locations shown

Strain gage recordings

Surface strains were recorded on each test specimen with three-element, 120-ohm rectangular rosette strain gages (KFG-3-120-D17-11L1M3S, Kyowa, Japan) attached to the geometric center of the prepared surface of each aluminum, polyurethane foam, and laminated polyurethane specimen using standard strain gage application procedures. This particular strain gage was chosen as a general purpose, pre-wired gage which has previously been used across several similar cortical strain studies in both composite and cadaveric bone models [6],[7],[22],[23],[28]. In femoral testing, three strain gages were applied to the medial aspect of each composite and cadaveric femur. Gages were placed at 25 mm increments beginning at 25 mm below the level of the lesser trochanter. The maximum and minimum principal strains at each rosette were recorded at peak loading using a strain conditioner/amplifier system (System 7000, Vishay Micro-Measurements, Raleigh, NC).

Digital image correlation

Three-dimensional DIC measurement techniques were employed in all test scenarios in this study. While two-dimensional DIC can be adequate with simple specimen geometries and loading conditions, three-dimensional DIC was chosen for all cases in this study for consistent comparisons in both simple and complex specimen geometries. A speckle pattern was applied to each specimen following strain gage application using off-the-shelf, commercial white and black spray paint. A white base was sprayed onto the surface of each specimen to form a thin, uniform background (and applied over the strain gage). After the base coat was dry, a speckle pattern of black was sprayed onto the same surface. Speckles were generated by manually applying partial pressure to the spray paint nozzle, inducing splatter instead of a mist spray. After some investigator practice, speckles can be applied in a repeatable density and size following this method. Paint splatter was manually adjusted until average speckle size and density matched manufacturer guidelines (Aramis 5M v.6.2.0, GOM, Inc., Braunschweig, Germany). Strain gages were not removed from the specimens prior to DIC speckling and measurement prep in order to reduce localized thinning or alteration of the specimen by sanding and scraping between tests.

The two CCD cameras (2448 × 2025 pixels) were also calibrated using the procedure described by the manufacturer, resulting in a calibrated measurement volume of 155 mm × 130 mm × 110 mm. The analysis input parameters in DIC were facet size, facet step and calculation base. The facet size was 15 pixels; increasing the facet size may improve the precision of point recognition without affecting the sensitivity to local strain variations (but increases calculation time). The 15-pixel facet size is consistent with the setting used in other biomechanical test set-ups [24]. A 13-pixel facet step and a computation size of 15 pixels were implemented. These two variables influence the precision and local accuracy of the calculated strains. Prior studies [17] have shown that when evaluating strains in the mid-area of a specimen (i.e., away from the edges), the average strain in the gage section of a specimen is independent of the facet size, facet step, and computation base.

Baseline images were taken of the unloaded samples and a series of five images were taken at peak loading to confirm the value of the reported strains at the location of the DIC strain measurement point. The overall size of the DIC measurement point in this analysis was 11.5 mm square, compared to the 10 mm base width of the physical strain gage. The maximum and minimum principal strains were recorded for the DIC method at the same location as the physical strain gages on each specimen.

Finite element modeling

To verify the accuracy of the strains recorded by the DIC and the strain gages, finite element (FE) models were generated to provide strains of the aluminum, polyurethane foam, and laminated polyurethane foam specimens. The models were created based on the average dimensions shown in Table 1. Loads and boundary conditions were simulated as described previously for each experiment using ANSYS finite element software (Version 13.0; Canonsburg, PA). Hexahedral elements were used for all FE models and convergence tests were completed successfully. The material properties are provided in Table 3.

Table 3. Finite element model material properties

Material	Poisson's ratio	Young's modulus (GPa)
Aluminum alloy	0.3	73.0
Polyurethane foam	0.3	0.64
Laminate layer	0.3	1.19

3. Results

A paired-samples t-test ($\alpha = 0.05$) was conducted to compare the maximum and minimum principal strains between DIC and strain gage values (Table 4) for each experiment. There was a significant difference ($p < 0.05$) in the maximum and minimum principal strain values for both compression and bending tests of the non-laminated polyurethane foam. No significant differences were found between the principal strains for compression and bending tests of the aluminum and laminated polyurethane foam. There was a significant difference ($p < 0.05$) in the principal strain values between the experimental strain values for each measurement technique and the numerical strain predictions (Fig. 3).

While both strain gage and DIC measurement techniques showed differences in strain values compared to the expected (numerical) results, the average percent difference across all measurements was 15.7% for the DIC method and 18.4% for the strain gage technique (Fig. 4). The measured strain values for the non-laminated polyurethane foam were overestimated (i.e., larger) by 17.6% for the DIC technique and underestimated (i.e., smaller) by 24.5% for the strain gage technique when compared to the numerical FE predictions.

Table 4. Principal strain measurements for the DIC and strain gage techniques

	Max principal strains (με)			Min principal strains (με)		
	DIC	Strain gage	*p*-value	DIC	Strain gage	*p*-value
Bending (Aluminum)	838 ± 34	821 ± 15	0.21	–287 ± 14	–273 ± 7	0.06
Bending (Foam)	926 ± 63*	679 ± 32*	<0.0001	–275 ± 28*	–168 ± 17*	<0.0001
Bending (Laminate)	710 ± 81	680 ± 69	0.16	–184 ± 25	–174 ± 5	0.05
Compression (Foam)	849 ± 122*	465 ± 63*	<0.0001	–2218 ± 340*	–1552 ± 295*	<0.0001
Compression (Laminate)	1131 ± 117	1089 ± 105	0.18	–3058 ± 185	–3018 ± 176	0.44

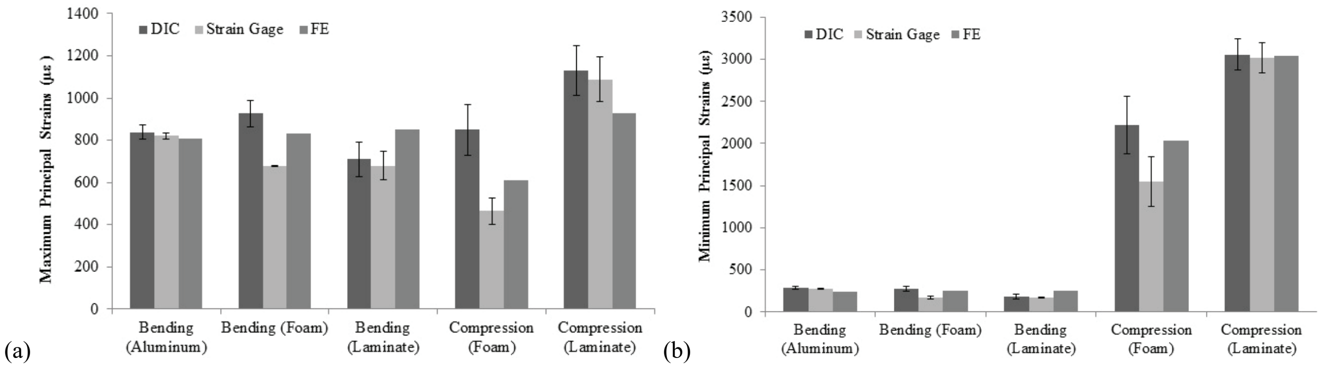

Fig. 3. Maximum (a) and minimum (b) principal strain values including standard error for each measurement technique and material under bending and compression loading. All measured experimental values (DIC and strain gage) demonstrated statistically significant differences from the theoretical values ($p < 0.05$)

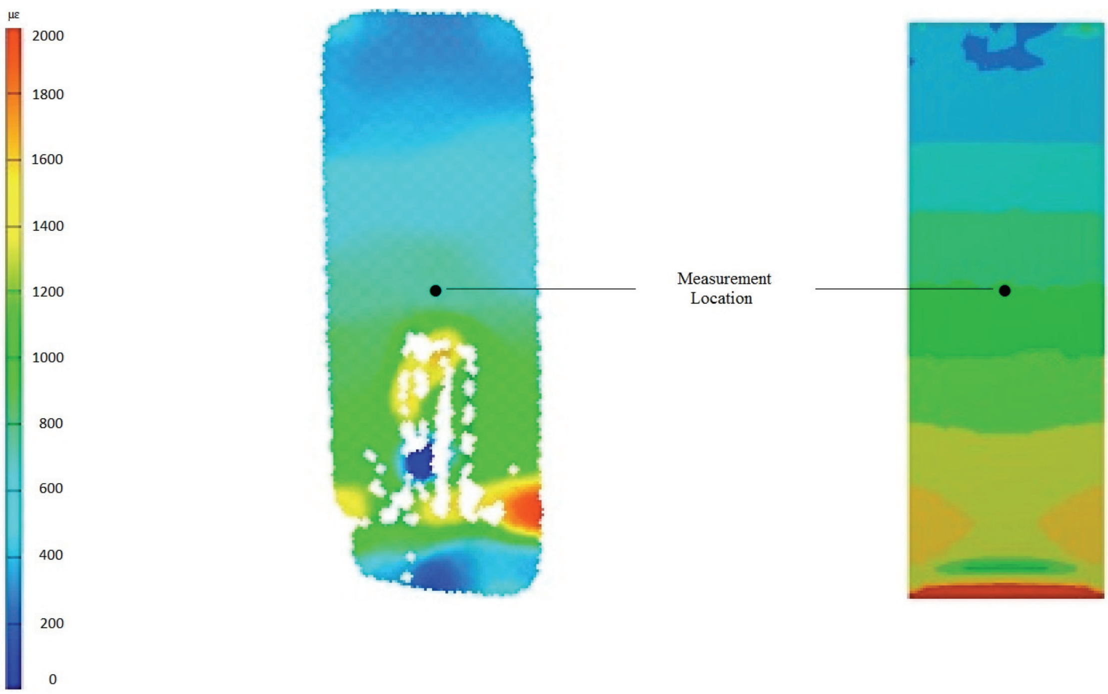

Fig. 4. Maximum principal strain distribution for a laminated foam block: (left) DIC technique; (right) FE results. (Note that the empty space in the DIC strain data coincides with strain gage wire placement that interferes with the speckle coating on the specimen)

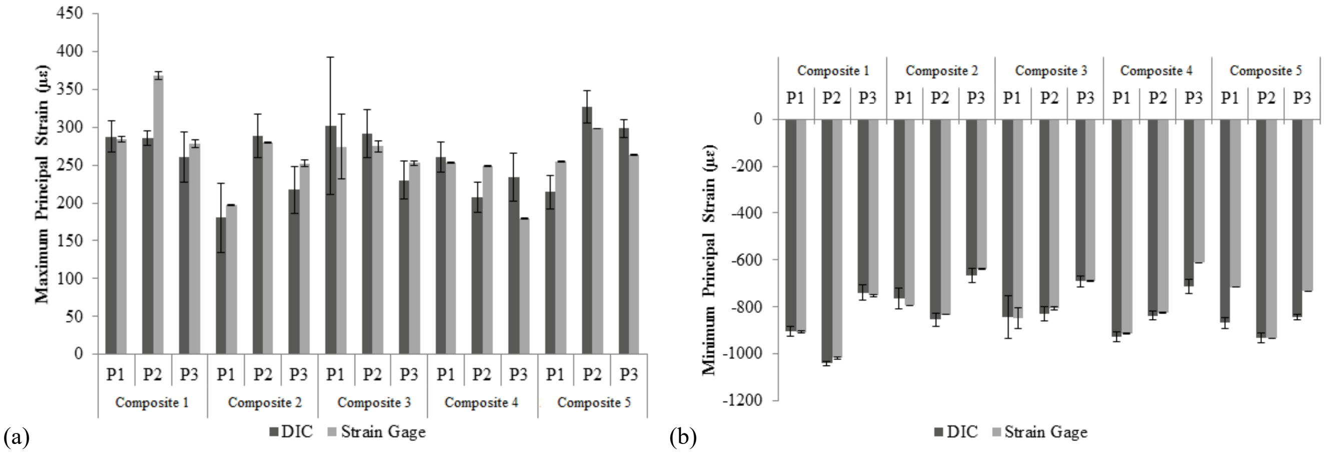

Fig. 5. Maximum (a) and minimum (b) principal strain values including standard error for composite femurs under simplified single-stance loading

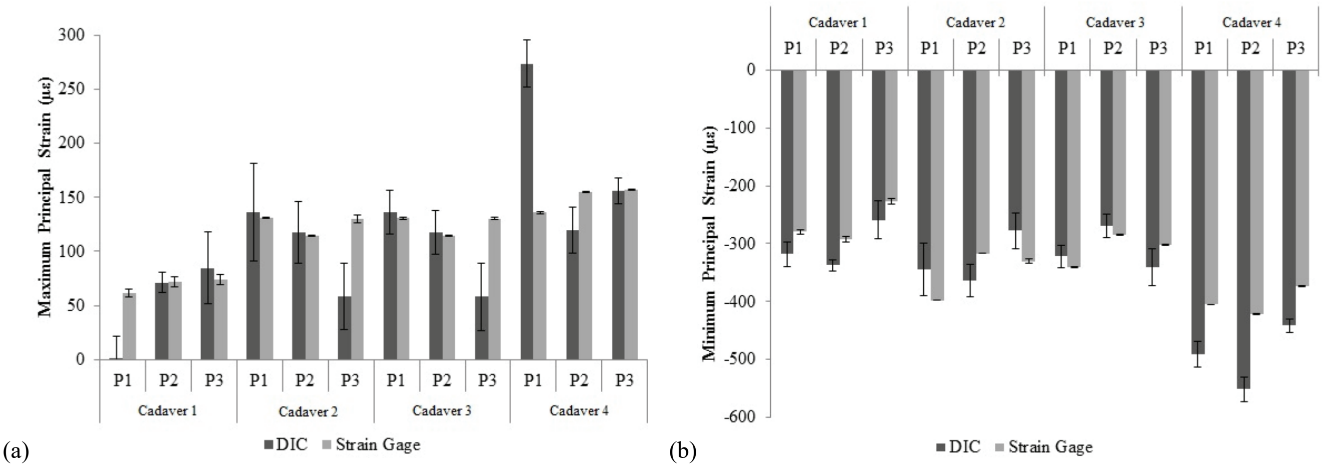

Fig. 6. Maximum (a) and minimum (b) principal strain values including standard error for cadaveric femurs under simplified single-stance loading

Table 5. Comparison of *p*-values at principal strain measurements for composite and cadaveric femurs

Composite femurs					Cadaveric femurs			
Bone	Position	*p*-value at max	*p*-value at min		Bone	Position	*p*-value at max	*p*-value at min
Composite 1	2	0.903	0.823		Cadaver 1	2	0.017	<0.001
	3	<0.001	0.304			3	0.910	0.808
	4	0.632	0.38			4	0.495	0.234
Composite 2	2	0.729	0.552		Cadaver 2	2	0.544	< 0.001
	3	0.764	0.386			3	0.928	0.244
	4	0.329	0.544			4	0.031	0.018
Composite 3	2	0.722	0.807		Cadaver 3	2	0.022	0.004
	3	0.652	0.139			3	0.637	0.909
	4	0.4	0.919			4	0.050	0.963
Composite 4	2	0.74	0.406		Cadaver 4	2	0.356	0.003
	3	0.114	0.592			3	0.631	0.012
	4	0.152	<0.001			4	0.962	0.047
Composite 5	2	0.134	0.005					
	3	0.258	0.993					
	4	0.049	<0.001					

For each composite and cadaveric femur, a paired-samples *t*-test was performed comparing principal strains recorded using strain gages and DIC ($\alpha = 0.05$) (Figs. 5 and 6). Statistically significant differences between principal strains were observed in 5 out of 30 trials in composite femurs and 10 out of 24 trials in cadaveric femur specimens (Table 5). In composite femurs, there was an average relative difference of 29 ± 20 με in maximum and 36 ± 46 με in minimum principal strains between DIC and strain gage measurements. Comparatively, the average relative difference between DIC and strain gage measurements across all cadaveric specimens was 74 ± 95 με in maximum and 52 ± 31 με in minimum principal strains.

A two-tailed *F*-test was conducted to evaluate the variance in the DIC and strain gage measurements. There were no significant differences between the variances of the DIC and strain gage measurements for the aluminum alloy, laminated foam, composite femur, and cadaveric femur specimens. There was a significant difference ($p < 0.05$) between the strain measurements for the non-laminated polyurethane foam specimens under both bending and compression loading.

4. Discussion

The goal of this study was to evaluate the utilization of DIC strain measurement techniques for investigations involving traditional biomechanical test materials (e.g., metals, foam blocks, composite, and cadaveric bone models). The DIC and strain gage measurements and the associated variances were not significantly different for the metal and laminated polyurethane foams, with an average difference between the two measurement techniques less than 4%. While strain measurements showed a statistical difference from the numerical predicted strains, the percent differences were similar between the DIC (14.8%) and the strain gage technique (14%) for the metal alloy and laminated foam specimens. These results indicate that DIC strain measurement techniques provide the same level of accuracy as current strain gage technology, while providing accurate full-field strain measurements on metals and laminated composite bones.

The DIC strain readings and the strain gage measurements for the non-laminated foam blocks were significantly different in both compression and bending. Additionally, deviations from the predicted response increased to 17.6% and 24.5% for the DIC and strain gage techniques, respectively. A significant difference in the variance was also noted for tests involving non-laminated foam blocks. One possible explanation for the significant difference between the DIC and the strain gage measurements for the non-laminated foam specimen might be the localized stiffening of the foam due to the strain gage adhesive. The porous foam structure is more susceptible to the stiffening effects of the adhesive and the stiffer response influences strain measurements. Based on these findings, a researcher who traditionally uses strain gages on non-laminated polyurethane foam blocks may improve the accuracy of reported strains by implementing a full-field DIC strain measurement technique or using a less porous substrate material.

We found good correlation between DIC and strain gage data throughout the majority of measurements taken during composite femur testing, with a 16.7% disagreement rate, similar to that reported in a prior comparison [29]. Subsequently, we observed significant divergence in the strain response of cadaveric specimens in 41.7% of the measurements taken. The divergence between strain gage and DIC data in the cadaveric specimens was expected, as the variability of strain gage measurements on natural hard tissues is well documented [30]. Factors including bone preservation, method of strain gage protection, bone temperature, and bone hydration all contribute to the variability strain gage measurement in cadaveric specimens [30] and are certain to play a role in the deviation between DIC and strain gage data observed in cadaveric, but not composite femur testing. Best practice preparation techniques [5] and the associated variability of strain gage measurements on natural bone [30] are well documented. Further guidelines are needed to establish best practices and to determine the associated factors driving measurement variability in DIC techniques in the cadaveric scenario.

Some prior studies utilize DIC techniques exclusively with non-physiological specimens [2], [4], [19], [29]. Väänänen et al. [29] published a systematic evaluation of repeatability of surface strain measurement in fourth generation composite femurs, as well as a comparison of surface geometry mapping in comparison with CT image data. Their study employed three-dimensional DIC measurement on the anterior aspect of six proximal femur models loaded compressively until femoral neck fracture. In this study, the authors reported interspecimen-variation in strain between identical composite femurs at equal loading between 20% and 25% and surface geometry agreement within 0.5 mm of CT imaging data across the majority of the bone surface [29]. Cadaveric bone specimens, however, offer increased complexity of measurement and procedural techniques due to irregular surface geometries, tissue perspiration and speckle adhesion [12].

Only a few studies have set out to directly evaluate the use of DIC in bone biomechanics. Gilchrist et al. [11] utilized DIC strain methods in comparison with a rosette strain gage applied to the femoral neck and applied a speckle pattern over the same measurement location. The authors reported a mean difference of 127 ± 239 με between strain gage and digital image correlation determined minor strains [11]. Comparisons between DIC and strain gage techniques were only made at a single measurement point per specimen [11]. In the current study, we observed a mean difference of 52 ± 31 με in minimum principal strains in the medial aspect of the implanted cadaveric femurs.

Notably, several other studies have used DIC in order to validate finite element models [9], [10], [13] on composite bones, while two other studies have used DIC to validate finite element models of cadaveric specimens [14], [15]. In general, these studies have resulted in similar strain responses between FE model predictions and DIC measurements.

Results of these studies must be interpreted with consideration of some important limitations. Depending on specimen porosity, surface finish, or moisture level, it may be difficult to apply the speckle coating to the test material. While most specimens require the application of a white base coat prior to speckling for adequate contrast, some types of specimens may only require black speckling. Producing a quality speckle pattern can be a variable art, and is a process which directly impacts the ability to capture accurate results. In the author's experience, a significant amount of practice is required to produce a repeatable and high-quality speckle field. Likewise, error can be introduced into DIC measurement by data fallout due to poor speckle quality or local surface abnormalities. However, DIC surface modifications may have less impact on the surface strains than application of a traditional strain gage, which can also be difficult in the case of biological specimens. One major advantage of the DIC technique is the ability to view a full strain field for a test specimen, while applying a thin paint layer onto the specimen as opposed to permanently adhering a strain gage with a stiff underlying adhesive.

5. Conclusions

In conclusion, digital image correlation (DIC) is a relatively new technique within the field of biomechanics which can be used to measure strain in bone and bone substitutes. The results of this investigation show agreement between the "gold standard" strain gage measurements and the DIC strain results, particularly for high density materials. The DIC strain results were also similar to strain gage measurements in the amount of deviation from numerical predictions which provides greater confidence in the application of DIC in future biomechanical analyses.

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