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Technical note

Flexible impulse transfer using a Newton's Cradle-inspired catheter: A feasibility study

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ABSTRACT

A major challenge during minimally invasive surgery is transfer of high forces through small, flexible instruments, such as needles and catheters, because of their low buckling resistance. In this study, we determined the feasibility of using a Newton's Cradle-inspired catheter (patented) to transfer high-force impulses. Exerting a high-force impulse on the tissue increases the critical buckling load and can prevent buckling. The system comprised an input plunger onto which the impulse is given, a (flexible) shaft filled with Ø2 mm stainless steel balls, and an output plunger to transfer the impulse to the target tissue. In the proof-of-principle experiment, the effect on efficiency of clearance (0.1, 0.2, and 0.3 mm), length (100, 200, and 300 mm), shaft type (rigid vs. flexible), curve angle (0, 45, 90, 135, and 180°), and curve radius (20, 40, 60, and 100 mm) was determined. The catheter delivered forces of 6 N without buckling. The average impulse efficiency of the system was 35%, which can be further increased by optimizing the design. This technology is promising for high-force delivery in miniature medical devices during minimally invasive surgery.

1. Introduction

Sir Isaac Newton developed the *Newton's Cradle* to demonstrate the conservation of momentum and energy (see Eq. (1)). *Newton's Cradle* consists of a series of identical metal balls suspended in a metal frame by two wires in a v-formation (Fig. 1). If one ball is pulled away from the other balls and is then released, it will gain momentum p (the product of its mass m (kg) times its velocity v (m/s)), which will subsequently be converted into an impulse J (the integral of the force F (N) over the time dt (s) for which it acts) when it strikes the second ball (see Eq. (1)). This second ball will acquire, and subsequently transfer, the impulse to the third ball, and so on, producing a pressure wave that propagates through the intermediate balls. When the pressure wave reaches the most distal ball, this ball gains momentum and reverses the direction of motion of the pressure wave. Assuming entirely elastic behavior, momentum loss is zero, enabling a successive oscillating motion of the two outer balls.

$$dp = m \cdot dv = \sum F \cdot dt \qquad (1)$$

with:

dp = change in momentum p (kgm/s)
m = mass (kg)
dv = change in velocity v (m/s)
F = force (N)
dt = time interval for which the force F acts (s)

A *Newton's Cradle*-inspired mechanism, such as that illustrated in Fig. 2, is interesting for medical applications in which high forces need to be transferred by flexible or ultrathin devices, such as guidewires, catheters, or needles, which are susceptible to buckling. Buckling is characterized by a sudden sideways deflection of the slender device subjected to high compressive loads; this is unwanted as it can cause damage to the surrounding tissues and decreases the force that can be applied by the tip, e.g., when puncturing a coronary occlusion in the heart using a guidewire. Exerting an impulse on the tissue enables high forces

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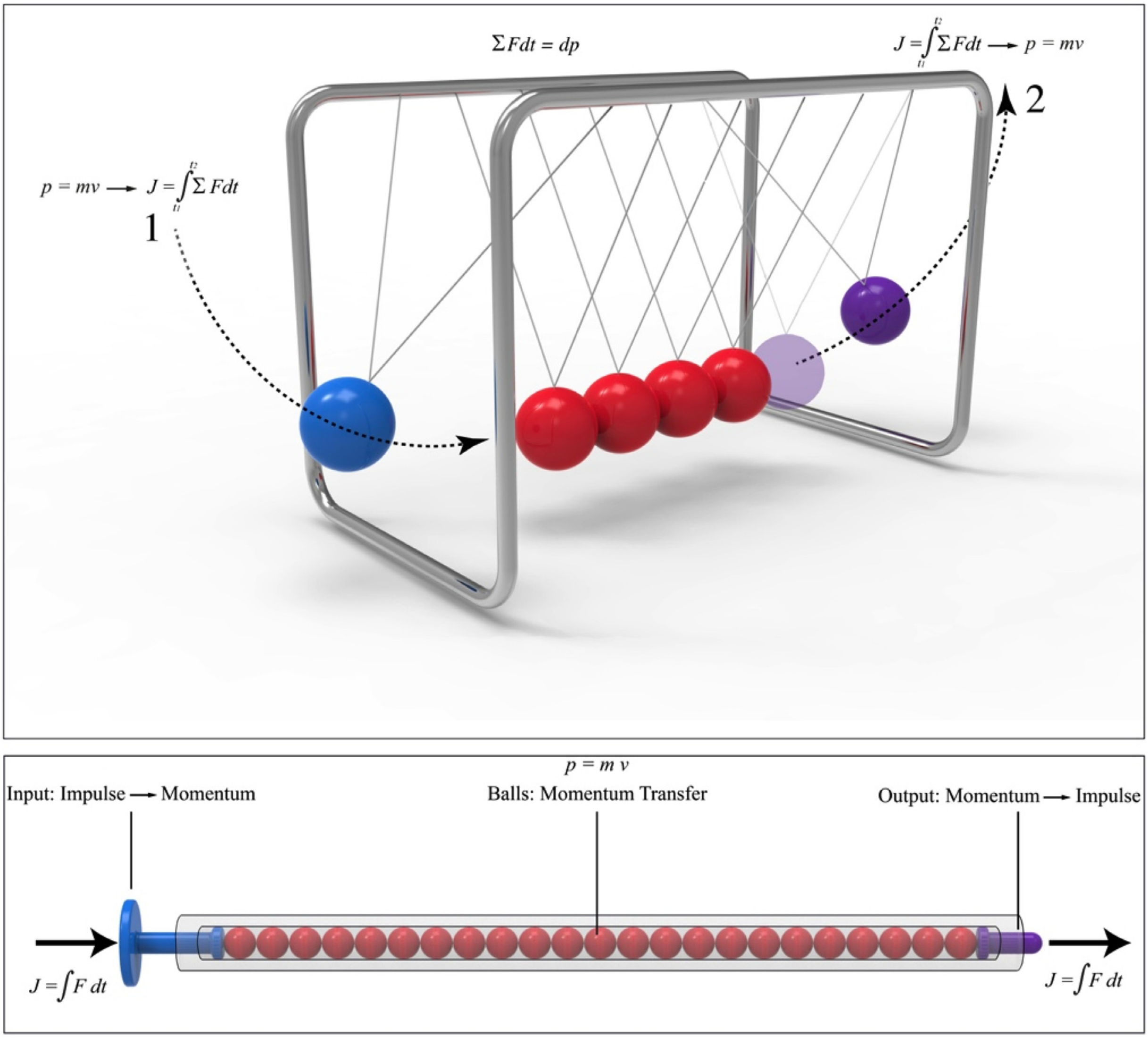

Fig. 1. *Newton's Cradle* mechanisms. Top: *Newton's Cradle*. Bottom: Schematic representation of the *Newton's Cradle*-inspired catheter. Color indications: blue = input ball, purple = output ball, and red = intermediate balls. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

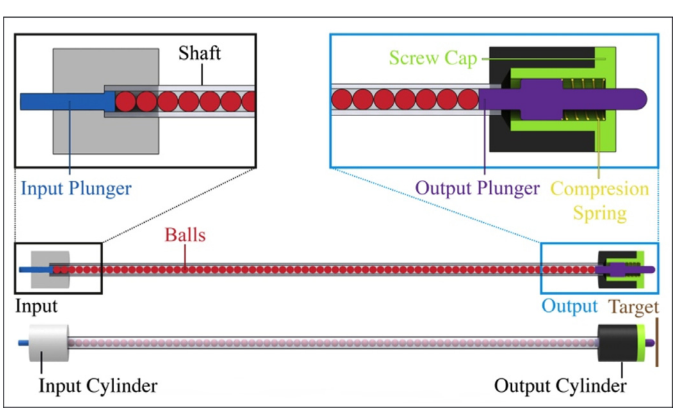

Fig. 2. Schematic illustration of the *Cradle* prototype for testing purposes. Color indications: black = output housing, blue = input plunger, green = output cap, grey = input housing, purple = output plunger, red = balls. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

to be applied on tissues as it prevents buckling in two main ways. (1) The use of an impulse increases the maximum compressive load, also known as the critical load, the device can withstand before buckling [1]. (2) Using an impulse to penetrate tissues can decrease the penetration load by minimizing energy losses to the environment. This effect can be mainly attributed to the inertia of the target tissue that acts as a counterforce to the incoming impulse, and thus decreases energy loss by tissue movement and deformation [2,3] (see also Sakes et al. [4] on this topic). In this study, we investigated the feasibility of using a *Newton's Cradle*-inspired mechanism for low-friction transfer of high-force impulses through slender shafts. For more information on the theory of *Newton's Cradle* concept, readers are directed to the studies Kinoshita et al. [5], Ceanga et al. [6], and Donahue et al. [7].

The Newton's Cradle inspired mechanism may need to conform to different requirements depending on the specific medical application. Here, we distinguish between two types of medical applications: (1) needle applications; and (2) catheter applications. In the case of needle applications, rigid, preferably hollow shafts are needed to administer medicine or take a biopsy. For catheter applications, on the other hand, a flexible shaft is required to travel through the vasculature from the incision point to the lesion site. The dimensions and flexibility of the shaft should conform to that of currently available instruments. A specific catheter application that would benefit from high-force delivery through a flexible shaft is in the treatment of chronically occluded coronary arteries, in which forces of over 1.5 N are needed for puncture.

2. Materials and methods

2.1. Experimental goal

The goal of our experiment was to explore the possibility of using a *Newton's Cradle* concept, and thus a series of spherical balls, to transfer an impulse through a long, slender shaft. To achieve this goal, the efficiency of the impulse transfer, and the capability of the *Cradle* prototype to exert high peak forces without buckling, needed to be determined. For comparison, a Ø2 mm stainless steel needle (wall thickness 0.2 mm, $L = 300$ mm) can withstand approximately 2.6 N before buckling whereas a similar size catheter can withstand only 0.003 N [8].

2.2. Cradle prototype

The *Cradle* prototype consisted of a Ø2.5 mm shaft filled with Ø2 mm stainless steel balls (Fig. 2). The balls were contained within a shaft with two polymer caps at either side. At the proximal and distal end were Ø1.5 mm plungers that transferred the input impulse to the balls and propagating pressure wave of the balls into an output impulse onto the target, respectively. A compression spring was placed distal to the output plunger to ensure contact between the balls. The bending stiffness of the *Cradle* prototype could be altered by changing the pre-tension between the balls. During insertion, the pre-tension can be removed to ensure the bending stiffness of the *Cradle* prototype did not exceed that of currently available support catheters to prevent blood vessel wall damage. Notably, the *Cradle* prototype is not torsion stiff.

2.3. Experimental variables

2.3.1. Dependent variables

To determine the quality of the impulse transfer, the following dependent variables were measured at a fixed input impulse:

(1) *The output impulse (J_o)* delivered by the output plunger.
(2) *The output peak force ($F_{peak\ out}$)* delivered by the output plunger.

2.3.2. Independent variables

The effect of the specific configuration of the *Cradle* prototype on impulse transfer was researched by sequentially altering the following independent variables:

(1) *Clearance (c).* The inner diameter of the shaft was varied to determine the influence of clearance between the balls and shaft. The clearance most likely affects the impulse and peak force efficiency through two main effects: (1) friction between the balls and the shaft, and (2) misalignment of the balls. The clearance between the balls and the shaft was set to 0.1, 0.2, and 0.3 mm.
(2) *Shaft type.* To determine the effect of the shaft material on the efficiency of the device, two different shafts were tested: a rigid stainless steel shaft and a double-braided axially and radially stiff cardiac catheter (*Mach 1^{tm}* 8F, Boston Scientific, Marlborough, MA, USA).
(3) *Length (L).* Energy is lost during each collision. The number of balls, and thus contact points, is likely to have an effect on impulse and peak force efficiency. The effect of the number of balls on energy loss was determined by setting the length of the shaft to 100, 200, and 300 mm, with a total of 50, 100, and 150 Ø2 mm balls in the shaft, respectively.
(4) *Curve angle (α).* Endovascular interventions cross several intersections and curves in the body. To determine the effect of these curves on energy loss, the prototype was subjected to single-constant radius curves ($r = 40$ mm) with four different curve angles (45, 90, 135, and 180°) and the impulse and peak force efficiency were compared with that in the straight configuration.
(5) *Curve radius (r).* The effect of curve radius was assessed by subjecting the prototype to single-constant radius curves with a curve angle $\alpha = 90°$ and four different curve radii: 20, 40, 60, and 100 mm.

The *input impulse (J_i)* and the associated *input peak force ($F_{peak\ in}$)* delivered onto the input plunger were set at 0.4 Ns and 5 N, respectively, and were measured throughout the experiments.

2.4. Experimental apparatus

An experimental prototype was built to determine the feasibility of the *Cradle* prototype (Fig. 3). The input impulse was delivered by a solenoid actuator connected to a construction rail and was set to approximately 0.4 Ns using a voltage of 5.5 V (Power unit *ES* 030-5, Delta Elektronika, Zierikzee, the Netherlands) and an actuation time to 100 ms. The delivered input and output impulses were measured using two miniature S-beam load cells (*LSB200 FSH00102* and *LSB200 FSH00104*; FUTEK Advanced Sensor Technology, Inc., Irvine, CA, USA) located at the input and output, respectively. Data were acquired through connecting an analogue signal conditioner (*CPJ RAIL*, SCAIME, Annemasse, France) and a data acquisition system with a sampling rate of 10 kHz (*NI USB-6211*, National Instruments Corporation, Austin, TX) to the load cells. The load cells were controlled through *LabVIEW 2016* (National Instruments Corporation, Austin, TX).

To test the effect of curve angle and radius on impulse and peak force efficiency and to confine the prototype shaft, three stacked transparent laser cut 3 mm thick poly (meythlmethacrylate) (PMMA) plates were used: (1) a solid bottom plate, (2) a middle plate containing the curvatures (see Fig. 3), and (3) a solid top plate, in between which the prototype was placed. These plates were connected to a breadboard (*MB3030/M*, Thorlabs, Newton, NJ, USA) using four aluminum rods and M6 socket head screws. The solenoid and the input module were placed on two parallel construction rails, so they could be easily translated left or right to accommodate the different laser cut curves in the PMMA plates. The position of the curves is different per configuration. The effect of the position of the curve was not researched in this study. Additionally, an unconstrained test was performed to give a clearer understanding of buckling of the catheter during high-force impulse transfer.

2.5. Experimental protocol

From the dependent variables, two efficiencies were calculated: (1) impulse efficiency and (2) peak force efficiency. First, the effect of clearance between the balls and shaft on impulse and peak force efficiency of the *Cradle* prototype was determined using the 200 mm rigid stainless steel shaft. These tests determined the clearance with the smallest impulse losses. Second, the effect of shaft type was determined using the stainless steel shaft and 200 mm catheter in the straight configuration with a clearance of 0.3 mm. This clearance was selected based on off-the-shelf catheter availability. Third, the effect of shaft length on efficiency was

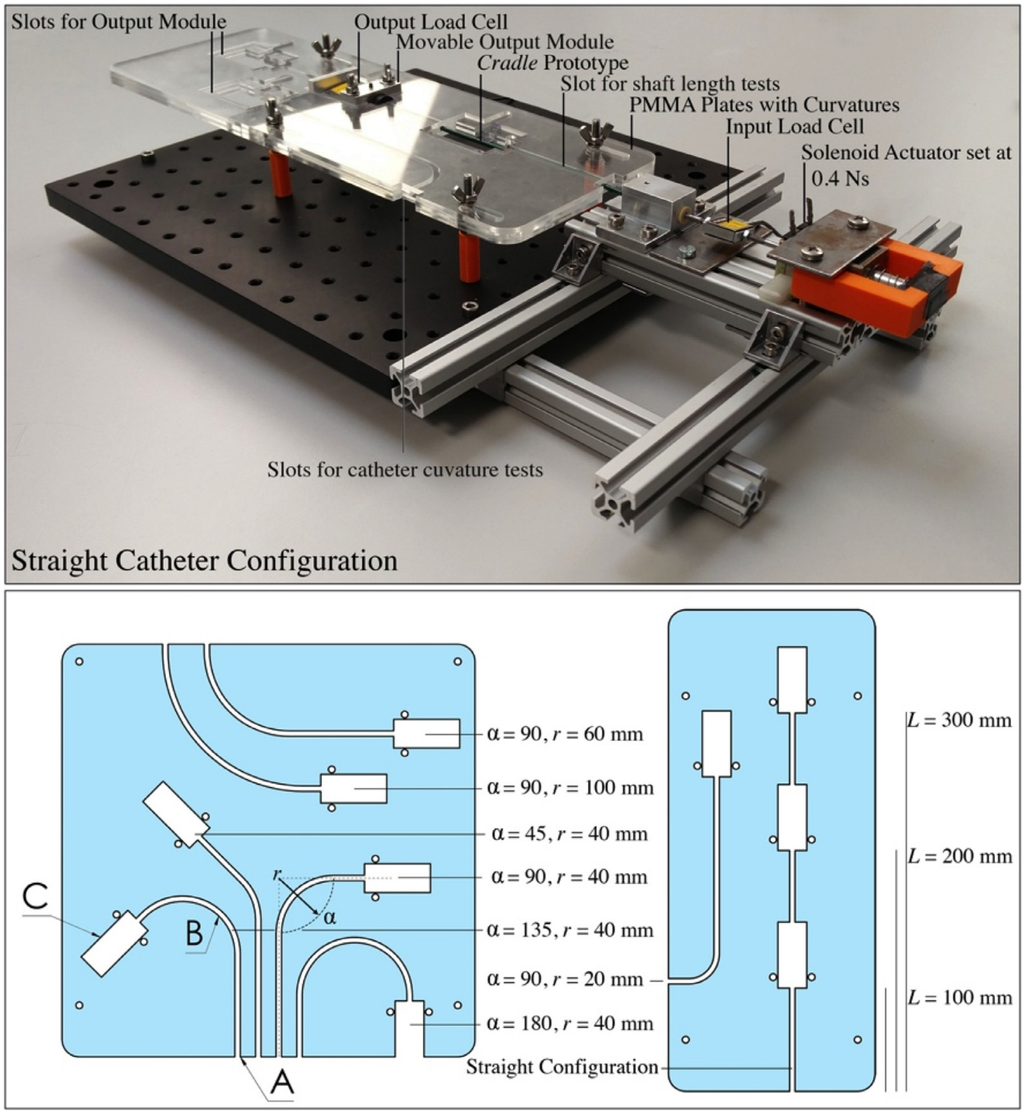

Fig. 3. Experimental configuration. Top: Complete experimental setup. Bottom: Schematic representation of the curvatures to which the *Cradle* prototype was subjected. Letter indications: α = curve angle (°), A= slot for catheter, B= curvature to which the catheter is subjected, C = insert for the output module containing the output load cell, L = length (mm) of the catheter, and r = curve radius (mm).

determined using the flexible catheter shaft in the straight configuration. Fourth, the effect of curve angle on efficiency was determined for the 200 mm flexible catheter shaft with a 40 mm curve radius. Fifth, the effect of curve radius on efficiency was tested for the 200 mm flexible catheter with a 90° curve angle. Each condition was tested 50 times. Finally, the buckling resistance of the prototype was evaluated for the 300 mm straight unconfined catheter shaft and a clearance of 0.3 mm. This condition was tested 3 times.

2.6. Data analysis

Data from the S-beam load cell were processed with *MATLAB 2015b* (The Mathworks, Inc., Natick, MA) to calculate the impact peak force and the impulses J_i and J_o (Ns) by integrating the force F (N) over the time dt (s) for which it acted. Mean output peak force and mean output impulse with the standard deviations were calculated per condition. The efficiency of the system (%) was determined in two ways: (1) dividing output impulse by input impulse and multiplying this value by 100, and (2) dividing output peak force by input peak force and multiplying this value by 100, for the different clearances, lengths, curve angles, and curve radii. One-way ANOVA was conducted to determine whether the clearance, length, curve angle, and curve radius had a significant influence on efficiency. A T-test was performed to determine the effect of shaft type.

3. Results

3.1. General findings

Fig. 4 illustrates an overview of the data from the load cells. The highest peak force measured during the experiment was 6 N in the straight configuration. As can be seen in Fig. 4, the load cells have a block shape with a slight peak at the start. The input peak force was not constant (see Table 1) for all tests even though the solenoid was calibrated and powered using the 5.5 V setting. Furthermore, the input impulse duration varied per configuration: impulse duration for the straight catheter configuration was approximately 70 ms, and for the rigid stainless steel tube was approximately 30 ms. Furthermore, the catheter slightly shifted

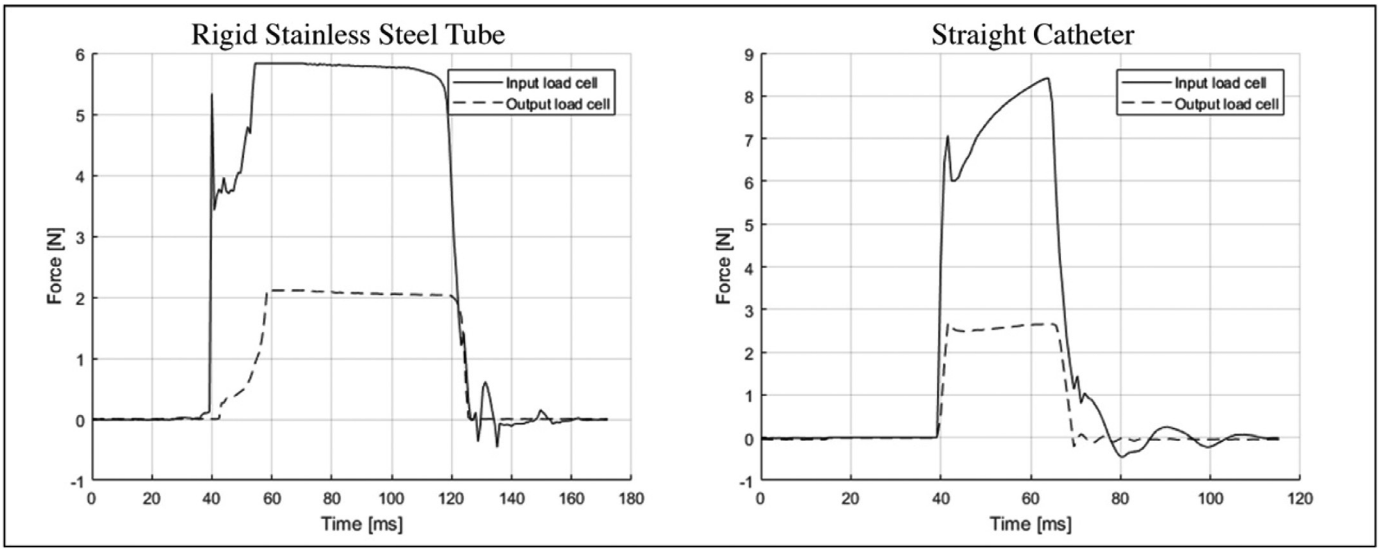

Fig. 4. Two examples of force versus time graphs of the input and output load cell for the straight configuration rigid stainless steel tube (left) and catheter (right).

Table 1
Effect of clearance and input impulse of the shaft on efficiency. The effect of clearance was tested using the rigid stainless steel shaft. The length of the capillary tube was set to 200 mm.

Clearance (mm)	Mean input peak force (N)	Mean input impulse (Ns)	Mean output peak force (N)	Mean output impulse (mNs)	Mean peak force efficiency (%)	Mean impulse efficiency (%)
0.1	9.9 ± 0.5	$2.4{\cdot}10^{-1} \pm 14.0{\cdot}10^{-3}$	3.7 ± 0.3	$8.8{\cdot}10^{-2} \pm 8.2{\cdot}10^{-3}$	36.9 ± 1.6	37.3 ± 2.0
0.2	4.0 ± 0.3	$1.5{\cdot}10^{-1} \pm 9.6{\cdot}10^{-3}$	2.3 ± 0.1	$8.6{\cdot}10^{-2} \pm 5.6{\cdot}10^{-3}$	57.1 ± 3.9	56.9 ± 3.6
0.3	7.5 ± 0.1	$1.7{\cdot}10^{-1} \pm 5.8{\cdot}10^{-3}$	2.4 ± 0.2	$5.4{\cdot}10^{-2} \pm 3.9{\cdot}10^{-3}$	32.0 ± 2.4	30.1 ± 2.1

Table 2
Effect of shaft type (Rigid stainless steel tube versus flexible catheter) and length on efficiency. The effect of shaft type was tested in the straight configuration with a clearance of 0.3 mm between the balls and the shaft.

Shaft type	Length (mm)	Mean input peak force (N)	Mean input impulse (Ns)	Mean output peak force (N)	Mean output impulse (mNs)	Mean peak force efficiency (%)	Mean impulse efficiency (%)
Rigid	200	7.5 ± 0.1	$1.7{\cdot}10^{-1} \pm 5.8{\cdot}10^{-3}$	2.4 ± 0.2	$5.4{\cdot}10^{-2} \pm 3.9{\cdot}10^{-3}$	32.0 ± 2.4	30.1 ± 2.1
Catheter	100	5.6 ± 0.1	$5.5{\cdot}10^{-1} \pm 16.3{\cdot}10^{-3}$	3.3 ± 0.08	$3.1{\cdot}10^{-1} \pm 9.9{\cdot}10^{-3}$	58.6 ± 1.0	60.0 ± 1.0
	200	4.9 ± 0.1	$4.9{\cdot}10^{-1} \pm 6.7{\cdot}10^{-3}$	1.8 ± 0.04	$1.7{\cdot}10^{-1} \pm 3.6{\cdot}10^{-3}$	36.6 ± 1.2	34.2 ± 0.8
	300	6.9 ± 0.2	$6.0{\cdot}10^{-1} \pm 16.1{\cdot}10^{-3}$	1.5 ± 0.06	$1.0{\cdot}10^{-1} \pm 7.1{\cdot}10^{-3}$	22.0 ± 1.1	17.3 ± 1.0

during actuation. Fig. 5 shows a graphical overview of the results. No buckling was observed in the unconstrained test, see Fig. 6.

3.2. Effect of clearance on efficiency

Table 1 presents an overview of the effect of clearance and input momentum on efficiency. As shown in Table 1, peak force efficiency ranged from 32% to 57% and impulse efficiency from 30% to 57%. Peak force and impulse efficiency were negatively, and statistically significantly, affected by clearance as determined by one-way ANOVA (F (2, 237) = 1796, $p = 6.6{\cdot}10^{-144}$ & F (2, 237) = 2067, $p = 10{\cdot}10^{-161}$, respectively). A clearance of 0.2 mm resulted in the highest efficiency.

3.3. Effect of shaft type and length on efficiency

Shaft type significantly influenced peak force and impulse efficiency of the system as determined by two t-tests (t (198) = 17.1, $p = 5.5{\cdot}10^{-41}$ & t (198) = 18.2, $p = 2.8{\cdot}10^{-44}$, respectively). The catheter in the straight configuration had a slightly higher efficiency. Impulse efficiency was 30.1 ± 2.1% for the rigid stainless steel tube and 34.2 ± 0.8% for the catheter (see Table 2). The length of the catheter negatively, and statistically significantly, affected peak force and impulse efficiency as determined by one-way ANOVA (F (2, 237) = 22,020, $p = 6.9{\cdot}10^{-270}$ & F (2, 237) = 33,933, $p = 4.8{\cdot}10^{-294}$).

3.4. Effect of curvature on efficiency

Curve angle negatively affected peak force and impulse efficiency of the system (see Table 3) as determined by one-way ANOVA (F(4, 395) = 106.6, $p = 1.2{\cdot}10^{-32}$ & F(4, 395) = 214.8, $p = 7.6{\cdot}10^{-72}$). Impulse efficiency was approximately 7.1% lower in the highly curved situation ($\alpha = 135°$ and $r = 40$ mm). Curve radius also statistically significantly affected efficiency, with the lowest impulse and peak force efficiency found for the lowest radius of 20 mm (F (3, 316) = 299.2, $p = 5.8{\cdot}10^{-92}$ & F (3, 316) = 380.8, $p = 1.5{\cdot}10^{-104}$).

4. Discussion

4.1. Summary of main findings

In this study, we determined the feasibility of using a series of balls to transfer an impulse through a flexible shaft. The catheter delivered high forces up to 6 N without buckling. Input impulse

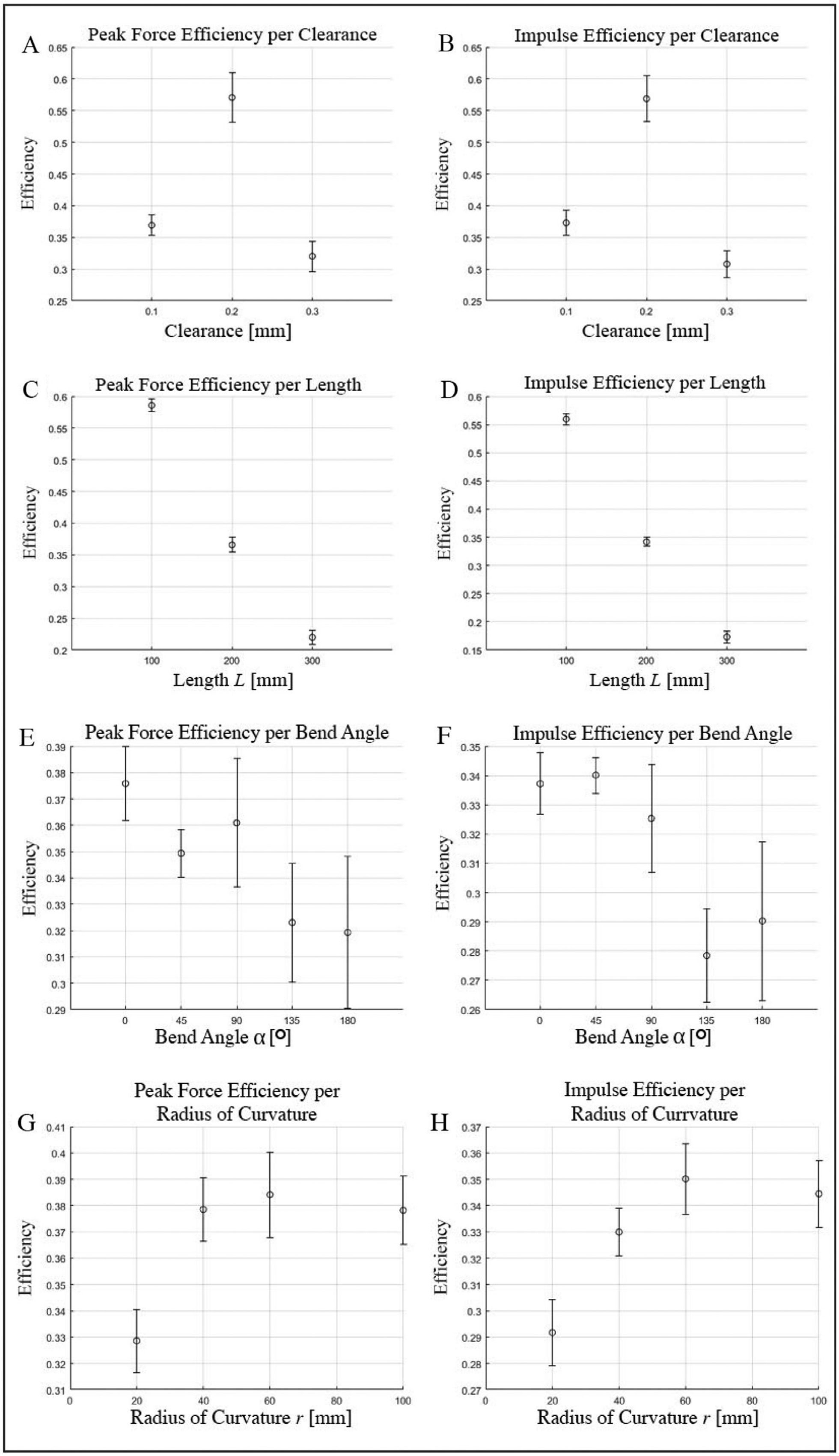

Fig. 5. Impulse and peak force efficiency results. (*A*+*B*) Peak force and Impulse efficiency for the different clearances of the rigid stainless steel shaft. (*C*+*D*) Peak force and Impulse efficiency for different catheter lengths. (*E*+*F*) Peak force and Impulse efficiency of the catheter in different curve angles ($r = 25$ mm). (*G*+*H*) Peak force and Impulse efficiency of the catheter in different curve radii ($\alpha = 90^\circ$).

Table 3
Effect of curvature of the shaft on efficiency. The effect of curvature was tested using the flexible catheter shaft with a clearance of 0.3 mm between the balls and shaft.

Shaft type	Curve angle (°)	Curve radius (mm)	Mean input peak force (N)	Mean input impulse (Ns)	Mean output peak force (N)	Mean output impulse (Ns)	Mean peak force efficiency (%)	Mean impulse efficiency (%)
Flexible	0	∞	4.9 ± 0.1	$4.9{\cdot}10^{-1} \pm 6.7{\cdot}10^{-3}$	1.8 ± 0.04	$1.7{\cdot}10^{-2} \pm 3.6{\cdot}10^{-3}$	36.6 ± 1.2	34.2 ± 0.8
Catheter	45	40	6.2 ± 0.1	$5.9{\cdot}10^{-1} \pm 9.6{\cdot}10^{-3}$	2.2 ± 0.04	$2.0{\cdot}10^{-1} \pm 3.7{\cdot}10^{-3}$	35.0 ± 0.9	34.0 ± 0.6
	90	20	3.1 ± 0.08	$3.3{\cdot}10^{-1} \pm 6.9{\cdot}10^{-3}$	1.0 ± 0.05	$9.5{\cdot}10^{-2} \pm 5.0{\cdot}10^{-3}$	32.9 ± 1.2	29.1 ± 1.3
		40	4.6 ± 0.07	$4.6{\cdot}10^{-1} \pm 6.2{\cdot}10^{-3}$	1.7 ± 0.05	$1.5{\cdot}10^{-1} \pm 3.9{\cdot}10^{-3}$	36.1 ± 2.5	32.5 ± 1.8
		60	3.6 ± 0.2	$3.7{\cdot}10^{-1} \pm 14.5{\cdot}10^{-3}$	1.4 ± 0.04	$1.3{\cdot}10^{-1} \pm 3.8{\cdot}10^{-3}$	38.4 ± 1.6	35.0 ± 1.3
		100	4.4 ± 0.1	$4.5{\cdot}10^{-1} \pm 10.9{\cdot}10^{-3}$	1.7 ± 0.06	$1.6{\cdot}10^{-1} \pm 5.5{\cdot}10^{-3}$	37.8 ± 1.3	34.5 ± 1.3
	135	40	5.6 ± 0.1	$5.4{\cdot}10^{-1} \pm 8.3{\cdot}10^{-3}$	1.8 ± 0.1	$1.5{\cdot}10^{-1} \pm 7.4{\cdot}10^{-3}$	32.3 ± 2.3	27.9 ± 1.6
	180	40	4.9 ± 0.1	$4.9{\cdot}10^{-1} \pm 10.9{\cdot}10^{-3}$	1.6 ± 0.1	$1.4{\cdot}10^{-1} \pm 13.4{\cdot}10^{-3}$	31.9 ± 2.9	29.0 ± 2.7

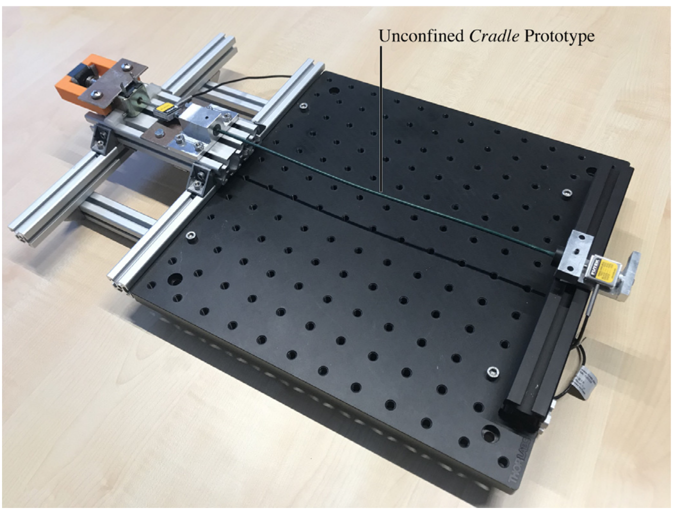

Fig. 6. Buckling resistance Proof-of-Principle Experiment with an unconfined *Cradle* Prototype. A catheter length of 300 mm and clearance of 0.3 mm were used. No buckling was observed in this experiment.

could not be increased in the current set up because of the capacity of the load cells; however, higher values can be achieved in future by increasing the input impulse. Impulse efficiency ranged between 17.3% and 60.0%, with an average of 35%. Length had the greatest influence on efficiency; curve angle and radii had minimal influence of 5–7%, which is beneficial for clinical situations where tortuous paths need to be followed towards the operation area. A slightly higher efficiency was found for the catheter shaft in the straight configuration than for the stainless steel tube, which is most likely due to a decrease in the friction coefficient between the balls and shaft. A clearance of 0.2 mm between the balls and shaft resulted in the highest efficiency. However, optimal clearance is yet to be determined and is likely to be 0.1 to 0.3 mm.

Input peak force was not constant throughout the tests even though the solenoid actuator was powered at an equal setting. This difference in input peak force was most likely caused by the difference in contact stiffness during impact in the different configurations, which in turn might be caused by the clearance, flexibility of the catheter shaft, or the curve. Also, the difference in contact stiffness changed the duration of the impulse for the different configurations, resulting in an increase of approximately 40 ms for the more flexible catheter configuration. The difference in magnitude and duration of the impulse may affect the efficiency of the system.

4.2. Limitations of this study

In this study we have not determined the feasibility of using this system in a clinical situation. An application that would benefit from a flexible tool that can deliver high forces without buckling is the endovascular treatment of Chronic Total Occlusions (CTOs; see also Sakes et al. [9] on this topic). Furthermore, as there is a clear trend towards even smaller, needle-like instruments, buckling will become a more important failure mode that needs to be considered in the near future. To prevent buckling and therefore increase the success rate of endovascular treatment of CTOs, as well as percutaneous needle procedures, a *Cradle*-inspired catheter could be beneficial.

4.3. Recommendations

The true working principle of *Newton's Cradle* is still under debate: there are multiple theories, none of which are completely correct. More research is necessary to determine the true nature of *Newton's Cradle*. Furthermore, in an effort to better understand the working principle of our system, research is required to accurately determine the energy losses due to friction and collisions in the shaft.

Optimization of the current design and prototype may increase efficiency. For this purpose, research should be conducted to determine the optimal clearance and material combination of the balls and shaft. It is also important to determine the optimal pre-tension, and thus contact stiffness, between the balls to minimize energy loss. If the pre-tension is insufficient, gaps may form between the balls, which will negatively affect the impulse transfer. Finally, it is also important to test the *Cradle* prototype in an environment that more closely resembles the clinical situation. The experimental configuration used in this study is rigid and does not account for the flexible environment encountered during minimally invasive surgery. The effect of the flexible environment on impulse transfer, and the effect of catheter movement during actuation, will need to be researched in a follow-up experiment.

To integrate the catheter in a laparoscopic or endovascular procedure requires single-handed control and adjustment of the output peak force. Device safety could be improved by incorporating the catheter shaft with the output plunger to minimize the loss of the indenter or the balls. Research is also required into actively steering the tip during the procedure as this can increase effectiveness and decrease the chance of tissue damage. Finally, to navigate through narrow tortuous pathways and decrease the invasiveness of the procedure, it is necessary to further reduce the outer diameter of the catheter. The design is simple, so diameter could be reduced towards a sub-millimeter scale. However, impulse and peak force efficiency will most likely decrease when miniaturizing the outer diameter, as the number of balls, and thus contact points, will increase. Increasing the length will also increase the required number of balls; therefore, other ball-shapes should be researched, such as cylinder-types, to minimize the number of contact points. Redesigning the shape of the balls could also make the *Cradle* prototype torsion stiff. For example, wedged-shaped elements having one degree of freedom could be used instead of spherical balls.

5. Conclusions

This study shows it is feasible to transfer high-force impulses through a series of balls confined within a rigid shaft or flexible catheter. This technique could be beneficial for medical instruments that are prone to buckling, such as guidewires and needles, particularly as there is a trend for further miniaturization of such instruments. More research is needed to enable high efficiency impulse transfer in smaller and longer devices. We will continue this investigation and will develop this tool into a handheld clinical prototype.

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Competing interests

None declared.

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Ethical approval

Not required.

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