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Pilot acute study of feedback-controlled retrograde peristalsis invoked by neural gastric electrical stimulation

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Abstract

Neural gastric electrical stimulation (NGES) is a new method for invoking gastric contractions under microprocessor control. However, optimization of this technique using feedback mechanisms to minimize power consumption and maximize effectiveness has been lacking. The present pilot study proposes a prototype feedback-controlled neural gastric electric stimulator for the treatment of obesity. Both force-based and inter-electrode impedance-based feedback neurostimulators were implemented and tested. Four mongrel dogs (2 M, 2 F, weight 14.9 ± 2.3 kg) underwent subserosal implantation of two-channel, 1 cm, bipolar electrode leads and two force transducers in the distal antrum. Two of the dogs were stimulated with a force feedback system utilizing the force transducers, and the other two animals were stimulated utilizing an inter-electrode impedance-based feedback system utilizing the proximal electrode leads. Both feedback systems were able to recognize erythromycin-driven contractions of the stomach and were capable of overriding them with NGES-invoked retrograde contractions which exceeded the magnitudes of the erythromycin-driven contractions by an average of 100.6 ± 33.5% in all animals. The NGES-invoked contractions blocked the erythromycin-driven contractions past the proximal electrode pair and induced temporary gastroparesis in the vicinity of the distal force transducer despite the continuing erythromycin infusion. The amplitudes of the erythromycin-invoked contractions in the vicinity of the proximal force transducer decreased abruptly by an average of 47.9 ± 6.3% in all four dogs after triggering-invoked retrograde contractions, regardless of the specific feedback-controlled
mechanism. The proposed technique could be helpful for retaining food longer in the stomach, thus inducing early satiety and diminishing food intake.

Keywords: gastric electrical stimulation, feedback control, obesity, weight control

1. Introduction

1.1. Gastric electrical stimulation

Gastric electrical stimulation (GES) is a method for electrically manipulating the lower stomach as a possible treatment for gastric motor dysfunction (Zhang and Chen 2006) by enhancing propulsive peristalsis (Forster et al 2003) or by producing retrograde peristalsis for the treatment of obesity (Sarna et al 1976). Various GES techniques have been proposed (Bortolotti 2002), including gastric pacing (Kelly and La Force 1980), low-energy high-frequency stimulation (Familoni et al 1997) and neural GES (Mintchev et al 1998, 2000, Neshev et al 2006).

Gastric pacing is a technique which stimulates the stomach at a frequency slightly higher than the natural electrophysiological frequency, entraining gastric electrical activity (GEA) (Kelly and La Force 1980) or the ‘slow waves’ (Sarna et al 1976). The slow waves represent a composite electrical phenomenon resulting from spontaneously occurring quasi-periodic cellular-level ionic exchange and are the necessary condition for gastric contractions to occur (Szurszewski 1998). The entrainment of the slow waves, therefore, has been suggested as an avenue to control gastric motility. However, the results associated with this technique have been modest, possibly due to the fact that intrinsic rather than extrinsic control remains the dominating factor determining the overall gastric motility index (De Luca et al 2004, Shikora 2004).

Low-energy high-frequency stimulation (Familoni et al 1997) does not entrain GEA but stimulates the stomach at frequencies significantly higher than the natural electrophysiological frequency. This method has shown some improvement of the mechanical activity in canine stomachs (Familoni et al 1997) but did not clearly demonstrate the ability to invoke contractions. The antiemetic effect associated with this technique has been reported when treating gastroparetic patients (Lin et al 2005), but the possible mechanisms for this phenomenon are yet to be verified and explained (Monnikes and van der Voort 2006).

In contrast to the previous methods, NGES generates multi-channel high-energy, high-frequency waveforms that can directly invoke contractions which can move gastric content in a controlled fashion depending on the synchronization between the stimulating channels (Mintchev et al 1998, 2000). NGES overrides any spontaneously existing electromechanical events and does not entrain the intrinsic gastric slow waves. By stimulating the local network of cholinergic neurotransmitters, repeated local contractions can be produced (Mintchev and Bowes 1997). This stimulation technique has been successful in accelerating gastric emptying of both liquids and solids (Mintchev et al 1998, 2000, Sobocki et al 2003) and in producing strong, externally controlled, retrograde contractions (Neshev et al 2006).

1.2. Obesity and gastric motility

The possibility to produce retrograde contractions in the stomach using GES is of particular interest for the treatment of obesity, which is considered as one of the most pressing health
problems of modern society. The prevalence of obesity has increased significantly in the period of 1988–1994 (Kuczmarski et al. 1997, Flegal et al. 1998) and is still growing (Flegal et al. 2002, Ogden et al. 2006). Various disorders and conditions have been related to obesity, many of which are life-threatening (Caterson et al. 2004). GES is a promising technique to treat obesity that could provide reliable long-term results without suffering from the side effects associated with pharmacotherapy (Bray and Greenway 1999) or from the postoperative complications related to anatomy-modifying bariatric surgery (See et al. 2002, Omalu et al. 2004). Invoked and appropriately controlled retrograde peristalsis could be an important avenue for delaying gastric emptying and thus indirectly controlling satiety and food intake (Chen 2004, Lee 2006). Since NGES is the only method that can invoke retrograde contractions, it could be regarded as the GES method of choice for the treatment of obesity. However, this technique is energy-demanding, and if utilized in an open-loop setup could pose difficult if not impossible long-term requirements for a multi-channel programmable implant. Moreover, recent chronic studies on experimental animals (Aelen et al. 2008) indicated that although the method was effective in reducing food intake, frequently invoked contractions in an open-loop system may lead to tissue accommodation resulting in NGES losing its ability to invoke contractility using the same amplitude of the stimulating voltage. Therefore, optimization of the invoked contractile patterns using feedback control is an important avenue to increase the effectiveness and the applicability of NGES by decreasing the energy demands for the stimulator and preventing gastric muscle fatigue due to over-stimulation.

Two feedback methods are proposed in this study. The first method is based on the mechanical nature of peristaltic contractions, which is detected by implanted strain gauges as changes in the physical shape of the gastric tissue in the vicinity of a contraction. The second method is based on the electrochemical nature of the contractions, which are detected as impedance changes in the vicinity of a contraction.

1.3. Tissue impedance

Tissue impedance has been routinely quantified as (Grimnes and Martinsen 2000)

\[ Z = \frac{zL}{A}, \]  

where \( Z \) is the electrical impedance of the tissue (\( \Omega \)), having both a resistive and a reactive (capacitive) component, \( z \) is its specific impedance (\( \Omega \) cm), \( L \) is the distance between the centers of the measuring electrodes (cm) and \( A \) is the tissue cross-sectional area (cm\(^2\)). In general, the tissue impedance between a transversely implanted electrode pair is frequency dependent, and this is manifested by the presence of a capacitive component. A model that is commonly used in the literature (Grimnes and Martinsen 2000) consists of a resistor in parallel with a non-ideal capacitor (figure 1). The resistive component represents the conductive characteristics of the extracellular body fluids, while the reactive component has its complex origin in the cellular membranes which act as leaky, non-ideal capacitors (Plonsey and Barr 1988).
1.4. Aim of the study

In order to control the timing of the stimulations and minimize the number of invoked contractions, effective NGES feedback methods are required. The purpose of the present pilot study is to demonstrate the feasibility of the mechanical and electrical impedance-based feedback methods in NGES.

2. Materials and methods

2.1. Experimental setup

Four healthy mongrel dogs (2 M, 2 F, weight 14.9 ± 2.3 kg) underwent general anesthesia induced with thiopental sodium (20 mg kg⁻¹ I.V. of Pentothal, Abbott Laboratories, Oakville, ON, Canada) and maintained with 1–3% isoflurane (Halocarbon Laboratories, River Edge, NJ, USA). One distal and one proximal set of stainless steel electrodes (Temporary Cardiac Pacing Wire, Weck Cardiac Pacing Wires, Research Triangle Park, NC, USA) were subserosally implanted at both sides of the anterior wall across the circumference of the distal antrum during laparotomy. Each set consisted of 1 cm active and reference electrodes, implanted at the opposite sides of the imaginary circumference perpendicular to the gastric axis. The distal set was placed 2–3 cm proximal to the pylorus and the proximal set was positioned 3–4 cm from the distal set in orad direction. In addition to the two electrode sets, two force transducers (Strain Gauge Transducer, RB Products, Stillwater, MN, USA) were implanted on the anterior gastric wall along the gastric axis, 3.5–4 cm apart, to capture gastric motility patterns (figure 2). Following the implantations, the abdomen was closed, and a custom-designed external neurostimulator and force transducer recorder were connected.

The experiments were divided in two separate groups. In a group of two dogs (1 M, 1 F), force transducer-based feedback was utilized (study 1) and in the group of the remaining two dogs (1 M, 1 F), inter-electrode impedance-based feedback was employed (study 2). In both study groups, the dogs were administered an intravenous infusion of erythromycin lactobionate (10 mg kg⁻¹, Abbott Laboratories, Saint-Laurent, QC, Canada) to drive the antral motor activity interoperatively. The erythromycin was diluted into 50 ml of 0.9% saline and was
infused over a period of 30 min (Otterson and Sarna 1990). Immediately after the start of the erythromycin infusion, the pharmacologically driven contractions were recorded for 15 min. After monitoring the pharmacologically driven propulsive contractions, the neurostimulator was utilized for 15 min to induce electrically driven retrograde contractions on ‘as needed’ basis using the two different feedback mechanisms. NGES was performed at voltage levels of 10 V peak-to-peak (Vpp), 100% duty cycle, 6 s on-time and 50% overlap according to a previous study (Aelen et al 2008) (figure 3). After the experiments, the dogs were euthanized with an intravenous injection of euthanyl, 480 mg/4.5 kg (Bimeda-MTC Animal Health Inc., Cambridge, Ontario, Canada). This research was approved by the Life and Environmental Sciences Animal Care Committee, University of Calgary, Calgary, Alberta, Canada.

2.2. Feedback neurostimulator design

Custom-designed hardware and software modules were developed to implement a prototype two-channel feedback-controlled neural gastric electrical simulator. The controlling software was designed using LabVIEW (National Instruments, Austin, TX, USA), which allowed substantial flexibility for making fast adjustments during the experiments. The neurostimulator generated two-channel, controlled, charge-balanced bipolar rectangular voltage waveforms at a frequency of 50 Hz with an adjustable amplitude, duty cycle, on- and off-time, and overlap between channels (see figure 3). By sequentially activating the channels, the direction of propagation could be controlled. The neurostimulator was implemented in software and the synthesized digital waveforms were converted to voltages by a DAQCard-AI-16XE-50 (National Instruments, Austin, TX, USA). A simple analog buffer amplifier provided the necessary current.

NGES for the treatment of obesity is based on the idea of overriding spontaneously existing slow waves by invoking local retrograde contractions to delay gastric emptying. However, spontaneously existing propulsive gastric motility is a rather infrequent process, only active in the period after a meal and in some of the four phases of the migrating motor complex (MMC) (Sarna 1985). Therefore, a feedback mechanism for NGES control could be very beneficial for retrieving the timing of the spontaneously existing contractile activity and adjusting the NGES timing accordingly. In the present study, the strength of gastric contractions, which is a measure of the spontaneously propulsive gastric motility, was assessed directly with the proximal force transducer (mechanical feedback) and indirectly with the inter-electrode impedance of the proximal implanted set of electrodes (impedance-based feedback) (see figure 2).

2.2.1. Mechanical feedback. To assess any mechanical changes in the physical shape of the gastric wall, two force transducers were utilized. Force measurements from these strain gauges
Figure 4. Block diagram of the overall feedback-controlled NGES technique including an impedance measurement system (IMS): (A) force-based feedback; (B) inter-electrode impedance-based feedback.

were acquired using a custom-designed analog bridge amplifier with frequency bandwidth between 0 and 1 Hz and were digitized using a 10 Hz sampling frequency (figure 4(A)). As a contraction travels toward a force transducer, the curvature of the gastric wall at the location of the transducer changes together with the strain exerted on the transducer gauge. Thus, a contraction can be detected by monitoring when the instantaneous force measured by the force transducer increases beyond a certain percentage of a predefined reference value. In the present study, this percentage was denoted as the trigger level. Once the force transducer reading increased beyond the trigger level, a stimulation session was issued.

However, since the curvature of the gastric wall constantly drifted even without the presence of contractions, a moving average (MA) of the force transducer reading over a certain period of time was employed as a reference curvature value. Thus, for a successful trigger event to occur, the force measured by the transducer had to increase quickly enough to not affect significantly the trigger level and be able to surpass it.

By adjusting the trigger level and the MA period, one can control the sensitivity of the triggering system. An increase in the trigger level implies a more stable triggering system since a much higher force reading is required to surpass the trigger level. On the other hand, an increase in the MA period decreases the stability of the triggering system, because the time required to alter the trigger level also increases, and a slower rise in the measured force by the transducer can reach the trigger level.

2.2.2. Electrical impedance-based feedback. It has been previously demonstrated that if electrical tissue properties are relatively constant over time, it is possible to estimate volumes
Figure 5. Synchronous demodulation. The current flowing through the inter-electrode impedance $Z$ driven by the voltage-controlled oscillator VCO is converted into voltage by the trans-impedance amplifier TIA, followed by analog processing and low-pass filtering (LPF).

encompassed by transversely positioned electrodes (Geddes and Baker 1989). Because antral contractions decrease the cross-sectional area at the location of a contraction and due to the specific electrode implantation technique that we utilized, we hypothesized that inter-electrode impedance could be a measure of contraction strength in the vicinity of a given transversely implanted electrode pair (see figure 2). With the transversely implanted electrodes in the proximal set located 4–5 cm apart (see figure 2), their inter-electrode impedance as a measure for the contractile strength at their vicinity can then be used in the feedback loop (figure 4(B)). The electrical impedance-based feedback method employs a stimulation triggering system, similar to the one that the mechanical feedback method uses. Once the instantaneous inter-electrode impedance changes beyond a certain percentage of the impedance’s MA over a predefined time period, a stimulation session is triggered.

2.2.3. Impedance measurement system. Figure 5 depicts the block diagram of the impedance measurement system (IMS). A voltage-controlled oscillator (VCO) generates a sinusoidal voltage waveform of adjustable frequency, which drives the inter-electrode tissue. The resulting current $I(t)$ flowing through the tissue is given by equation (2):

$$I(t) = \frac{V_0}{|Z(t)|} \sin(2\pi ft - \phi(t)) = \frac{V_0}{|Z(t)|} \left[\sin(2\pi ft) \cos(\phi(t)) - \cos(2\pi ft) \sin(\phi(t))\right],$$

where $V_0$ and $f$ are the amplitude and frequency of the generated voltage waveform, respectively; $|Z(t)|$ and $\phi(t)$ are the magnitude and phase of the inter-electrode impedance, respectively. It can be clearly seen that the resulting current $I(t)$ is amplitude-modulated by the inter-electrode impedance. The current waveform is converted to a voltage waveform via a trans-impedance amplifier (TIA). To recover the phase and magnitude of the inter-electrode impedance, synchronous demodulation is performed by multiplying the signal, respectively, with $\sin(2\pi ft)$ and $\cos(2\pi ft)$, followed by low-pass filtering (LPF). This is demonstrated by equations (3) and (4), where $K$ is a constant specific to the circuit components, and $Q$ is the current-to-voltage converter scale factor:

$$LPF\{QI(t) \cos(2\pi ft)\} = -\frac{K \sin(\phi(t))}{|Z(t)|},$$

$$LPF\{QI(t) \sin(2\pi ft)\} = \frac{K \cos(\phi(t))}{|Z(t)|}.$$
Assuming that $K$ is known, these expressions are solved by the developed software to determine the phase and magnitude of the inter-electrode impedance.

The frequency at which this inter-electrode impedance would be meaningful as a feedback parameter is the one at which the impedance difference between no contraction and maximum contraction is the highest. This frequency had to be determined experimentally because of the unique electrode configuration utilized in the study. Therefore, the impedance measurement system had to be frequency-adjustable, with a minimum of about 5 kHz, to ensure that no other bioelectric signals like ECG, EEG and EMG influenced the impedance measurements and to prevent stimulation of muscles and nerves (Baker 1989), and with a maximum of about 500 kHz, limited by the parasitic capacitances between the electrode leads as well as these between the ground and the measured object (Scharfetter et al 1999).

2.2.4. IMS calibration. The IMS has to be calibrated in order to account for the parasitic electrode impedance effect on the measured inter-electrode impedance. One can model the electrodes as a two-port network with unknown transmittance parameters that are to be determined through the calibration process (figure 6). Considering this electrode model, the measured impedance $Z_m$ is a function of the transmittance parameters $A$, $B$, $C$ and $D$, and the actual inter-electrode tissue impedance $Z$:

$$Z_m = \frac{AZ + B}{CZ + D}.$$  

The calibration procedure was carried out by connecting the known reference impedance to the electrodes and recording the measured value. This was performed 30 times for reference impedances with magnitudes in the range of 300–3000 $\Omega$, and phases in the range of 0 to $-\pi/2$ radians. From the collected data, the electrode transmittance parameters were calculated by least-squares estimation techniques (Wolberg 2006). Once the transmittance parameters are known, equation (4) can be solved for the inter-electrode tissue impedance.

3. Results

3.1. Pharmacologically driven and invoked contractions

The magnitudes of the electrically invoked contractions exceeded the magnitudes of the erythromycin-driven contractions (recorded when the stimulator was off) as measured by the two force transducers by 139.1 ± 38.9% at FT1 (the proximal force transducer) and by 62.1 ± 27.8% at FT2 (the distal force transducer) in all animals, an average increase of 100.6 ± 33.5%. The mean magnitude of the invoked contractions measured by FT1 was 1.34 ± 0.13 times greater than the one measured by FT2. Altogether, 208 occasions of invoked contractions and 284 occasions of pharmacologically invoked contractions were recorded by
both force transducers from all four dogs, regardless of the feedback-controlled mechanism utilized. A typical contractility pattern is presented in figure 7.

After the stimulator was turned on and invoked retrograde contractions were triggered, the mean amplitude of the pharmacologically driven contractions decreased by $47.9 \pm 6.3\%$ at FT1 in all animals regardless of the feedback mechanism utilized. Altogether, FT1 recorded 142 pharmacologically driven contractions from all four dogs prior to triggering invoked retrograde contractions (i.e. before turning on the stimulator) and 108 pharmacologically driven contractions were recorded after this triggering. After the application of the electrically invoked contractions, FT2 could not register any substantial number (>30) of erythromycin-driven contractions distinguishable from the overall baseline noise and respiratory artifacts, and therefore, per cent comparison was considered statistically not viable (see figure 7).

3.2. Mechanical feedback

The mechanical feedback method proved itself successful in effectively controlling the NGES timing. When an erythromycin-driven contraction registered by the proximal force transducer (FT1) reached the stimulation trigger level, the neurostimulator was turned on and produced a retrograde contraction starting at the distal channel. Once the measured force by the proximal force transducer FT1 reached the threshold due to the Ery-driven contraction, NGES was activated (see figure 7). Approximately 4 s later, the invoked contraction was detected by the distal force transducer FT2, because FT2 was located proximally to the distal electrode set. For both dogs, the frequency of the erythromycin-driven contractions significantly diminished immediately after the first NGES-invoked retrograde contraction was administered, a phenomenon that could be regarded as NGES-invoked gastroparesis.

![Figure 7. Erythromycin-driven and -invoked contractions. The force measured by the force transducers is in relative units (ru), since the static strain experienced by the transducer depends on uncontrollable factors related to positioning, suturing, etc. The threshold level for the proximal force transducer (FT1) is marked by a horizontal dotted line. The moment of NGES triggering is denoted with a dotted vertical line.](image-url)
Figure 8. Invoked gastroparesis. While the NGES system was off, Ery-driven contractions were not blocked and they managed to reach the distal force transducer (FT2). Once the NGES system was activated (at the 210th s), Ery-driven contractions were blocked and were detected only by the proximal force transducer (FT1). The retrograde nature of the contractions can clearly be seen as the contractions started in the distal channel and propagated to the proximal channel.

Remarkably, no erythromycin-driven contraction could reach FT2 once the NGES unit was turned on.

3.3. Electrical impedance-based feedback

3.3.1. Tissue interrogation frequency. The impedance measurement device was tested on a piece of beef tripe to find the optimal interrogation frequency. Two electrodes were placed on the tripe 5 cm apart and a frequency sweep from 5 kHz to 500 kHz was performed. After the first sweep, the tripe was pulled together so that the inter-electrode distance was reduced to 2.5 cm. Another frequency sweep was performed to find the impedance in this contracted state. The results from both sweeps are depicted in figure 9. There was no detectable change in the phase of the impedance. The inter-electrode impedance difference between contracted and non-contracted states was approximately 350 $\Omega$ for all frequencies, indicating there was no favorable frequency. Finally, a frequency of 50 kHz was chosen for the actual experiment, which was considered a preferred frequency for single-frequency bioimpedance measurements (Zarowitz et al 1993, Kon et al 2005).

3.3.2. Stimulator control. As expected, it was observed that as a contraction approaches the measuring set of electrodes (the proximal set), the magnitude of the inter-electrode tissue impedance decreased; however, the phase remained unchanged at approximately $-6$ degrees. Thirty measurements of the inter-electrode impedance were performed on each of the two dogs resulting in an average value of 910 $\Omega \pm 24$ $\Omega$ for the first dog and 1350 $\Omega \pm 35$ $\Omega$ for the second dog. The deviations resulting from erythromycin-driven and NGES-induced contractions were in the 100 $\Omega$ order of magnitude in both animals. Figure 10 depicts the
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**Figure 9.** Search for the optimal tissue interrogation frequency: 1—impedance magnitude of the tissue for 5 cm electrode spacing; 2—impedance magnitude of the tissue for 2.5 cm electrode spacing; 3—difference between curves 1 and 2. The approximately constant difference between curves 1 and 2 indicates no favorable interrogation frequency.

**Figure 10.** Impedance-based feedback. As a contraction passed the proximal electrode set, the inter-electrode impedance decreased. The inter-electrode impedance lagged the proximal force transducer reading (FT1) by approximately 5 s, because FT1 was implanted more proximally than the proximal electrode set. Once the inter-electrode impedance dropped below the threshold, the distal set, followed by the proximal electrode set, issued stimulation trains, and the resulting contractions were measured by FT2.

Results for the inter-electrode impedance-based feedback system. Once the inter-electrode impedance fell below the threshold, the distal channel began stimulating. After 3 s, the proximal channel also issued a stimulation train. Both invoked contractions were detected.
by the distal force transducer. The fact that the distal force transducer did not detect the erythromycin-driven contractions manifested the ability of the feedback method to trigger stimulation at the right time and effectively induce retrograde peristalsis while blocking the antegrade propagation of the pharmacologically driven propulsive contractions. Similar to the results with force-based feedback, pharmacologically driven contractions could not reach the distal force transducer and invoked gastroparesis was also observed in both dogs after the inter-electrode impedance-based feedback was activated.

4. Discussion

In the present pilot canine study the feasibility of two different feedback methods for controlling neural GES was investigated. A force transducer-based feedback method and inter-electrode impedance-based feedback method were both able to measure erythromycin-driven gastric contractions and to control the timing of stimulation in order to invoke retrograde contractility leading to temporary gastroparesis during a continuing erythromycin infusion. These two different feedback mechanisms were utilized to improve the effectiveness of retrograde neural GES for the treatment of obesity. The feasibility of both methods to control the timing of the neurostimulator in a very similar and comparable fashion was demonstrated in acute experiments. However, to optimize the feedback NGES, the effect of stimulation and feedback parameters must be further investigated in chronic animal models. Further work must be undertaken also to fully evaluate the performance of the proposed feedback stimulator and optimize the parameters of the feedback-stimulation protocol.

A previous chronic animal study of NGES for the treatment of obesity (Aelen et al 2008) showed adaptation of the gastric smooth muscle when stimulated with the same voltage amplitude level over several days. With the feedback loop closed, stimulation would only be applied in the presence of a natural contraction of pre-determined strength, instead of continuously stimulating (and fatiguing) the smooth muscle. These fewer NGES-invoked contractions would probably make it less likely to cause muscle adaptation to the applied voltage stimulation levels. Further, if muscle adaptation still occurs over a longer period of time, the voltage amplitude of stimulation can be adjusted to higher levels by measuring the amplitude of the invoked contractions with the feedback mechanism itself. In addition to the reduced chance of muscle adaptation by this more efficient way of stimulation, the demands on the batteries of the implant would be significantly lower. This brings the possibility of performing long-term chronic animal and human studies a step closer, which is necessary for converting this method into a practical clinical modality. Another energy-saving possibility could be the reduction of the amplitude of stimulation. In the present study, the amplitudes of the invoked retrograde contractions were higher than those of the erythromycin-driven contractions. Therefore, the voltage amplitude of stimulation could probably be lowered to produce contractions of similar strength to the natural contractions. This in turn would have a positive effect on the energy demands for a future implantable neurostimulator, would further reduce the muscle adaptation problem and diminish the refractory time so that the state of invoked gastroparesis could be achieved faster and more efficiently.

In a future implantable gastric neurostimulator, an inter-electrode impedance-based feedback system should be preferred over a force-based feedback setup in order to keep the surgical procedures minimally invasive and to reduce the technological requirements to the device. Further, the inter-electrode impedance-based feedback employs the same electrodes utilized for stimulation and therefore no extra wires have to be implanted. Only a minimal hardware enhancement of an implantable NGES device would be required to practically implement such a system.
The invoked gastroparesis that was observed during this study shows the potential of the method to reduce food intake by delaying gastric emptying. The reduced frequency of spontaneously existing gastric contractions disturbed on demand by the NGES-invoked retrograde contractions would be the main factor contributing to delaying gastric emptying which in turn could provoke early satiety and reduced food intake.

5. Conclusion

In this comparative pilot study of two different feedback methods for controlling neural GES, the inter-electrode impedance-based feedback emerged as a preferred technique to embed in an autonomous implantable neurostimulator.

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